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TITLE: Joint Loads and Cartilage Stress in Intact Joints of Military Transtibial Amputees:
Enhancing Quality of Life

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14. ABSTRACT The goals of this project are to identify exercises that maintain intact limb knee and hip joint and cartilage tissue loads at safe levels and, consequently, enhance the quality of life (QoL) via prevention of intact limb joint arthritis for military transtibial amputees. Progress includes the following. 1) Completed experiments for all participants (10 control and 10 amputee) in gait, cycling, and elliptical training exercises. 2) Analyses of knee angles correct for soft tissue artifact and crosstalk errors in 4 control and 4 amputee participants. 3) Completed analyses for all participants of ground reaction, pedal reaction, and resultant hip and knee joint loads. 4) Continues EMG-driven Opensim inverse dynamic analyses of knee joint contact loads and finite element analyses of cartilage tissue loads. During year 3, results and findings have been presented as eight conference papers and 5 MS theses; preparation of 2 journal papers has commenced.					
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1. INTRODUCTION

Abnormal biomechanics is a risk factor for knee joint osteoarthritis (OA) and is likely related to joint pain and OA among transtibial amputees. Although lower limb joint loading during gait for patients with high risk for OA has been examined, there are a lack of studies on joint loading for other fitness sustainment exercises and on predicting cartilage tissue loads of lower limb prosthesis users. The goals were to identify exercises that maintain intact limb knee and hip joint and cartilage tissue loads at safe levels and, consequently, enhance the quality of life (QoL) via prevention of intact limb joint OA for military transtibial amputees. The overall hypothesis was that intact limb joint biomechanics would depend on disability status (amputee vs. control) and exercise type (walking vs. cycling vs. elliptical training). Specific Aim 1 was to conduct motion analysis experiments and calculate intact limb knee and hip joint loads in amputee and control group subjects. Specific Aim 2 was to develop and use subject-specific finite element models to calculate intact limb knee joint cartilage loads in amputee and control group subjects. The results suggested that in order to reduce OA risk among transtibial amputees, the preferred rehabilitation and lifelong fitness sustainment exercises, in order of preference, are cycling, elliptical training, and walking. The results provided new biomechanical data that suggest that non-weight bearing exercises, such as cycling, should be emphasized as a critical component of an exercise regimen for transtibial amputees.

2. KEYWORDS

Amputee, prosthesis, knee, hip, biomechanics, cartilage, arthritis, finite element analysis.

3. ACCOMPLISHMENTS

What were the major goals of the project?

Objective 1: Calculate knee and hip joint loads for walking, cycling, and elliptical training exercises.

Task 1.1: Recruitment of n=10 amputee and n=10 control group subjects. Target date: 1/13/2018. Completion %: 100.

Task 1.2: Motion analysis experiments with amputee and control group subjects. Target date: 1/13/2018. Completion %: 100.

Task 1.3: EMG-driven inverse dynamic (ID) analyses with OpenSim for amputee and control group subjects. Target date: 1/13/2019. Completion %: 100.

Task 1.4: Statistical analyses of knee and hip joint contact resultants for amputee and control group subjects. Target date: 1/13/2019. Completion %: 100.

Objective 2: Calculate knee joint cartilage tissue loads for walking, cycling, and elliptical training exercises.

Task 2.1: Conduct MRI scans of intact limb knee joints of n=3 amputee and n=6 control group subjects. Target date: 1/13/2018. Completion %: 100.

Task 2.2: Develop solid models of the intact limb knee joints for n=3 amputee and n=3 control group subjects. Target date: 1/13/2019. Completion %: 100.

Task 2.3: Develop FE models of the intact limb knee joints for n=3 amputee and n=3 control group subjects. Target date: 1/13/2019. Completion %: 100.

Task 2.4: Conduct FE model simulations for amputee and control group subjects. Target date: 1/13/2019. Completion %: 100.*

Task 2.5: Statistical analyses of knee joint cartilage tissue loads for amputee and control group subjects. Target date: 1/13/2019. Completion %: 100.*

** A 100% completion rate is reported as all FE model simulations were conducted. However, some FEA model simulations did not achieve convergence at maximum targeted knee angles or loads and, thus, not all FEA model simulations were successful. For more details, see the material in Section 3 titled “Calculate knee joint cartilage tissue loads for walking, cycling, and elliptical training exercises”.*

What was accomplished under these goals?

The Appendix contains 1) a Bibliography for citations presented below and 2) select manuscripts that the reader is referred to for more details (1 journal paper, 2 internal technical reports, 1 MS thesis not yet published on Cal Poly’s Digital Commons public repository).

Experiments, analysis of ground and pedal reaction forces, and inverse dynamic analyses of resultant knee and hip loads.

Methods and results for this section were presented/published in 5 conference papers (Gutierrez-Franco et al., 2017; Heyde et al., 2018; Orekhov et al., 2017, 2018; Pottinger et al., 2018), 1 MS thesis (Orekhov, 2018), and 1 journal paper (Orekhov et al., 2019). **For more details, see the journal paper in the appendix (Orekhov et al., 2019) and the published MS thesis (digitalcommons.calpoly.edu/theses/2010).**

Specific objectives.

The objectives were to conduct motion analysis experiments and compare knee and hip biomechanics in gait, cycling, and elliptical training in order to identify appropriate exercises for transtibial amputees whom are at high risk for knee OA. The hypothesis was that knee and hip kinetics and kinematics would differ due to participant status (amputee, control), exercise (gait, cycling, elliptical), and leg type (amputee intact-residual, control dominant-non-dominant).

Major activities.

(1) Conducted gait, cycling, and elliptical training experiments with 10 transtibial amputee and 10 control participants. (2) Measured ground and pedal reaction forces for gait and machine exercises, respectively. (3) Performed inverse dynamic analyses to obtain hip and knee joint resultant loads. (4) Compared hip and knee joint kinetics, measured reaction forces, and knee flexion angles between exercises, between participant groups, and between intact-residual or dominant-non-dominant legs.

Significant results.

For the knee (see (Orekhov et al., 2019) in appendix), maximum compressive force was higher in gait vs. cycling for both amputee and controls. Maximum extension torque was higher in gait vs. cycling for amputee-intact and control non-dominant knees and higher in the intact vs. residual knee in gait for amputees. Maximum abduction torque was higher in gait vs. cycling for both amputee and controls. Amputee maximum knee flexion angle was higher in the residual vs. intact knee in gait and elliptical. Gait midstance knee flexion angle timing was asymmetrical for amputees.

For the hip (see (Orekhov, 2018) at digitalcommons.calpoly.edu/theses/2010), maximum compressive force was highest in gait and lowest in cycling for both amputee and controls. Maximum extension torque was lowest in cycling for both amputee and controls. Maximum abduction torque was highest in gait for both amputee and controls and, for amputee dominant hips, lower in cycling vs. elliptical.

The results suggest that cycling, and likely other non-weight bearing exercises, may be preferred exercises for amputees due to significant reductions in biomechanical asymmetries and joint loads.

Inverse dynamic analyses of knee and hip joint contact compressive loads using OpenSim. Methods and results for this section were presented/published in 3 conference papers (Fernandez et al., 2019; Mavrommati et al., 2017; Pottinger et al., 2017), and 1 MS thesis (Pottinger, 2018). 1 journal paper is in preparation. **For more details, see the internal technical report in the appendix (Fernandez et al., 2020) and the published MS thesis (digitalcommons.calpoly.edu/theses/1988).**

Specific objectives.

The objective was to calculate and compare knee and hip joint contact compressive forces for gait, cycling, and elliptical training exercises that may be considered for rehabilitation and lifelong fitness sustainment guidelines that consider minimizing OA risk. The hypotheses of this study were three-fold: knee and hip joint contact compressive forces would differ due to (H1) exercises for the amputee group (amputee gait vs. amputee cycling vs. amputee elliptical) and (H2) exercises for control group (control gait vs. control cycling vs. control elliptical) and (H3) participant status for each same exercise (amputee vs. control gait, amputee vs. control cycling, and amputee vs. control elliptical).

Major activities.

(1) Conducted OpenSim inverse dynamic analyses to calculate knee and hip joint contact compressive forces for transtibial amputee and control participants in gait, cycling, and elliptical training exercises. (2) Compared results between exercises and between participant groups for each exercise.

Significant results.

See (Fernandez et al., 2020) in the appendix. For maximum knee and hip joint contact compressive forces, significant differences due to exercise type were found only within a particular population, supporting H1 for amputees and H2 for controls. Specifically, knee and hip joint forces were consistently highest in gait and lowest in cycling for both control and amputee groups (gait>elliptical, elliptical>cycling, gait>cycling). No differences were found across amputee-intact and control-dominant legs for each exercise (e.g., amputee-intact gait vs. control-dominant gait). Thus, the results did not support H3.

The results suggest that, for TT amputees, cycling should be emphasized over gait and elliptical training for rehabilitation and/or lifelong fitness sustainment exercises due to lower joint contact loads that may minimize OA risk. This study's results also suggest that, for TT amputees, elliptical training should be preferred over gait.

Instrumenting stationary bicycle with seat post and handle bar load cells for validation of OpenSim pelvic residual loads.

Methods and results for this section were presented/published in 1 conference paper (Wash et al., 2019) and 1 MS thesis (Wash, 2019). **For more details, see the published MS thesis (digitalcommons.calpoly.edu/theses/2015).**

Specific objectives.

Although OpenSim was used to calculate knee and hip joint contact loads as described in the previous section, OpenSim was developed for gait analysis where the only external loads applied to the body are ground reaction forces applied to the feet. Also, OpenSim uses an optimization algorithm that creates and minimizes (relative to a zero value) a pelvic residual force vector in order to correct for experimental errors. However, in cycling there are external loads applied to the body by the handlebars and seat that create a substantial resultant at the pelvis and it is unknown if the minimized pelvic residual forces will be similar to the resultant produced by the handlebar and seat loads. Thus, the objective of this study was to verify that OpenSim's Residual Reduction Algorithm (RRA) with a recent OpenSim model for high knee flexion is capable of appropriately accounting for experimentally measured handlebar and seat forces during seated cycling through the addition of pelvic residual forces. If successful, this study would suggest that OpenSim can be used to obtain accurate estimates of knee and hip joint contact loads during cycling.

Major activities.

1) Designed, fabricated, installed, and tested a custom handlebar stem and seat post each instrumented with 6-axis load cells (ATMI, MA, USA). 2) Completed experiments with 5 participants. 3) Used the measured load cell data to calculate handle bar & seat equivalent (HBSE) loads at the pelvis. 4) Compared the HBSEs to the pelvic residuals used in OpenSim's RRA.

Significant results.

See (Wash, 2019) at digitalcommons.calpoly.edu/theses/2015. Graphical comparisons of the RRA and HBSE results showed strong correlations in Anterior- Posterior (A-P) and Superior-Inferior (S-I) force directions and to a lesser degree, Medial-Lateral (M-L) force and S-I moment directions. M-L and A-P moment plots showed the least correlation between RRA and HBSE. Statistical comparisons showed RRA errors were likely within 5.2% body weight (BW) for forces and 2.4% BW*height for moments. Considering the average participant height and weight of 167.7 cm and 63.6 kg, respectively, recommended error ranges for RRA are roughly $\pm 4.0\%$ BW for forces and $\pm 7.2\%$ BW*height for moments.

The results suggest that OpenSim is an appropriate analysis tool to use for calculating hip and knee joint loads during cycling.

Analysis of knee angles using algorithms to correct for soft tissue artifact (STA) errors.

Methods and results for this section were presented/published in 5 conference papers (Skaro et al., 2017, 2018; Tucker et al., 2018b, 2018a, 2019), and 1 MS thesis (Skaro, 2018). 1 journal paper is in review and 1 journal paper is in preparation. **For more details, see the internal technical report in the appendix (Tucker et al., 2020) and the published MS thesis (digitalcommons.calpoly.edu/theses/1922).**

Specific objectives.

Transfemoral (TF) amputees have increased risk of intact knee osteoarthritis (OA) that may be influenced by abnormal kinematics. Previous studies have measured knee angles for TF amputees in gait and cycling but none have corrected for soft tissue artifact (STA), a primary source of error using skin- based marker systems. Thus, the objective of this study was to develop and implement two STA correction algorithms to obtain knee angles for amputee and control participants in gait and cycling. The hypothesis was that intact (amputee) and dominant (control) knee angles would differ in gait but not cycling between TF amputee and control participants.

Major activities.

1) Developed an experimental protocol using clusters of 7 motion analysis markers placed on the thigh and shank. 2) Include those marker clusters on all experiments with amputee and control participants described above. 3) Developed custom MATLAB codes (MathWorks, MA, USA) for 2 state-of-the-art analytical techniques to reduce STA-induced errors in measured anatomical knee angles during motion analysis. The first analytical technique modeled each sub-cluster of 3 markers as a triangular Cosserat point element (TCPE) (a TCPE is a body that experiences only homogeneous deformations). Using parameters which can measure deformation and relative motion, we estimated the true underlying rigid body motion of the body segments. To validate our calculated knee angles, the second analytical technique used a more traditional STA correction method (i.e. the Procrustes Solution [PS] method). For both methods, a floating axis joint coordinate system was used to define flexion/extension (FE), internal/external (IE) rotation, and adduction/abduction (AA) angles. 4) Developed a state-of-the-art analytical technique to reduce crosstalk errors in measured anatomical knee angles during motion analysis using a statistical technique employing linear algebra call Principal Component Analysis (PCA) (see (Skaro, 2018) at digitalcommons.calpoly.edu/theses/1922 for more details).

Significant results.

See (Tucker et al., 2020) in the appendix for more details. There existed no differences for FE, IE, and AA angles due to subject status (control vs. TF) or exercise type (gait vs. cycling) (Fig. 1). Further analysis of the data revealed that there existed a difference ($p=0.006$) between the timing of the minimum adduction angle which occurred at ~50% and 66% of the gait cycle for TF and control participants, respectively.

The results suggest that differences in kinematics between TF amputee-intact and control-dominant knees may be found in metrics other than maximum angles (e.g. the timing and asymmetry of flexion angles in both mid-stance and swing phases) as reported in (Orekhov et al., 2019) and discussed above. Thus, future work should include investigating the timing and range of 3-D knee angles that are corrected for STA errors using the TCPE method.

Calculate knee joint cartilage tissue loads for walking, cycling, and elliptical training exercises. Methods and results for this section were presented/published in 3 conference papers (Lane et al., 2018; Rumery et al., 2018a, 2018b) and 3 MS theses (Lane, 2018; Rumery, 2018; Stearns, 2020). **For more details, see the published MS thesis (Lane, 2018)**

(digitalcommons.calpoly.edu/theses/1833) and the MS thesis (Stearns, 2020) in the appendix.

Specific objectives.

We hypothesized that tibial cartilage contact pressure (CCP) depends on an individual's amputee status and exercise type. The objectives were to 1) develop 6 (n=3 control, n=3 amputee) subject-specific FEA models from MRI scans, 2) calculate CCP in gait and cycling exercises using knee resultant loads and knee angles from each subject but imposed only on one robust “generic” FEA model, and 3) calculate CCP in gait, cycling, and elliptical exercises using knee contact loads and knee angles from each subject using all 6 subject-specific FEA models.

Major activities.

1) Developed all 6 solid models from the MRI scans. 2) Developed all 6 FEA models. 3) Post-processed input data for knee joint loads and angles for all 6 models for gait, cycling and elliptical. 4) Validate FEA models using published experimental data. 5) Conduct FEA model simulations using one robust generic model with gait and cycling knee resultant loads and angles for 3 control and 3 amputee participants to determine trends in medial and lateral contact pressures between groups and exercise types. 6) Conduct FEA model simulations using gait, cycling, and elliptical knee contact loads and angles for 3 control and 3 amputee participants to determine trends in medial and lateral contact pressures between subject groups and exercise types. Due to the complexity of the methods, the reader is referred to the published MS thesis (Lane, 2018) (digitalcommons.calpoly.edu/theses/1833) for more details on objectives 1-4 to the MS thesis (Stearns, 2020) in the appendix for more details on objectives 1-4 and 6.

Significant results – objective 5.

For all 6 subjects, dominant (control) or intact (amputee) knee compressive and shear forces, moments, and flexion angles corresponding to the maximum compressive resultant force in gait and cycling were applied to the generic FEA model. Two-way ANOVA determined tibial CCP differences due to amputee status and exercise type in the medial and lateral compartments, followed by Tukey pairwise comparisons when significant (defined by $p < 0.05$).

For each subject, medial CCP was higher than lateral CCP for both gait and cycling (which agrees with many previous studies that found higher loading in the medial compartment) (Figure 1). In the medial compartment, there was no significant difference in medial CCP due to exercise ($p=0.088$), amputation status ($p=0.760$), or the interaction of the two ($p=0.959$). In the lateral compartment, there was no significant difference in lateral CCP due to amputation status for gait ($p=1.000$) or cycling ($p=0.997$). However, lateral CCP was lower in cycling vs. gait for both amputee ($p < 0.001$) and control ($p < 0.001$) subjects.

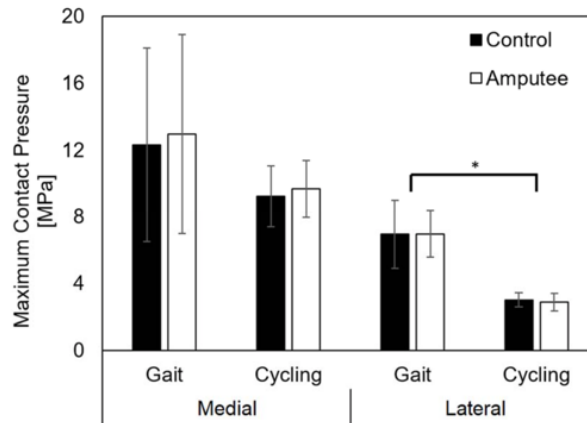


Figure 1. Maximum medial and lateral cartilage contact pressures for control (n=3) and amputees (n=3) subjects in gait and cycling. * = lateral CCPs were lower in cycling vs. gait for control and amputee subjects.

Lateral compartment tibial CCP was lower in cycling vs. gait likely due to lower varus-valgus moments that were measured experimentally. The lower varus-valgus moments in cycling likely leads to reduced loading in the lateral compartment and increased loading it in the medial compartment. **Results indicate that regardless of subject type, cycling exercises may decrease CCP and, thus, reduce OA risk in the lateral compartment. However, these conclusions must be treated with caution because only resultant knee loads were used.**

Significant results – objective 6.

As noted above, the conclusions presented for Objective 5 must be treated with caution because only resultant knee loads were used. To obtain a more accurate FEA simulation of knee joint CCP, knee joint contact loads were used in this aim. More specifically, the muscle-driven inverse dynamic analyses conducted for Task 1.3 of Aim 1 partition the joint resultant loads into muscle and contact loads. Since the contact loads are the actual loads applied to the cartilage surfaces, in this aim we conducted FEA simulations with applied contact loads.

For all 6 subjects, dominant (control) or intact (amputee) knee compressive contact force and flexion angles corresponding to the maximum compressive contact force in gait, cycling, and elliptical training were applied to each of the 6 subject-specific FEA models. However, FEA model convergence to the 100% compressive contact force was obtained in gait, cycling, and elliptical for only 1 control and 2 amputee subjects; thus, only those results are presented. It is noted that the FEA model simulations solved the boundary-value problem in an incremental fashion with increasing values of 0-100% contact force; each FEA model simulation that was not successful failed to converge at some value of contact force that was less than 100% of the targeted experimental value.

CCP results were normalized by body weight and are reported here in units of MPa/N. For the 1 control subject in gait, the medial CCP was 0.018 and the lateral CCP was 0.021. For the 2 amputee subjects in gait, the medial CCPs were 0.028 and 0.024 and the lateral CCPs were 0.029 and 0.036.

For the 1 control subject in cycling, the medial CCP was 0.018 and the lateral CCP was 0.010. For the 2 amputee subjects in cycling, the medial CCPs were 0.018 and 0.017 and the lateral CCPs were 0.014 and 0.013.

It was not possible to conduct a statistical analysis of these results due to a n=1 for control subject results. However, the results do suggest several notable trends.

In gait, amputee subject medial CCPs were 33-56% higher than for the control subject and amputee subject lateral CCPs were 14-38% higher than for the control subject. **Thus, there was a trend toward higher normalized CCPs in amputees vs. control for gait.**

In cycling, amputee subject medial CCPs were 0-6% lower than for the control subject and amputee subject lateral CCPs were 30-40% higher than for the control subject. Although there was a trend toward higher normalized CCPs in the lateral compartment in amputees for cycling, the CCPs in amputees in cycling were substantially less than for gait. More specifically, **amputee subject medial CCPs were 29-36% lower in cycling vs. gait and lateral CCPs were 52-64% lower in cycling vs. gait. Thus, these results reinforce conclusions from Aim 1 that cycling, but not gait, may be a preferred exercise for transtibial amputees.**

Stated goals not met.

As noted in Section 3, select simulations did not achieve convergence at maximum targeted knee angles or loads and, thus, not all FE model simulations were successful. Originally, we planned to develop 6 (n=3 control, n=3 amputee) subject-specific FEA models from MRI scans and conduct FEA model simulations of CCP using calculated joint contact forces in gait, cycling, and elliptical training at times of highest compressive load. This was a highly novel aim, as only 1 previous study had attempted a similar aim (Klodowski et al., *Multibody System Dynamics*, 37:271-290, 2017). However, that previous study only developed a FEA model for 1 subject and only considered gait. Generally, we found that it was very challenging to 1) develop successful FEA model simulations for all 6 participants and 2) obtain FEA convergence for the relatively high knee flexion angles observed in cycling and elliptical exercises. Ultimately, we had more success with modeling cycling because, although the knee flexion angle was highest in cycling, the contact loads were substantially less than for gait and elliptical training. The combination of high flexion angles and high contact loads for elliptical training likely led to the result that FE model convergence failed for most of those models. Although we ran all 6 subject-specific models for all load cases, most of those FEA models did not converge. Thus, here we report only the results from the successful FEA model simulations.

Other achievements.

Funded students Greg Orekhov and Elizabeth Heyde received the 1st place award in the BS level student paper competition at the 2018 World Congress of Biomechanics (sponsored by the Bioengineering Division, American Society of Mechanical Engineers).

What opportunities for training and professional development has the project provided?

Training activities.

Students and Faculty completed on-line Citi Program training (course: Biomedical Researcher: Basic/Refresher) on protection of human subjects in research. Students that were hired completed extensive training via 11 training modules (marker placement, equipment calibration, gait experiment, cycling & elliptical experiments, Citi Program certification, EMG use, data post-processing, OpenSim, CPR, review of research & safety protocol documents, statistical analysis) over a 10-week period.

Faculty mentors consistently mentored students on general research skills, communication skills, graduate school and career choices, resume development, ethics in human subjects research, and diversity and inclusivity issues.

Professional development.

Students submitted and presented 17 conference papers at major academic conferences where they developed their communication and networking skills.

Students led weekly HMB Lab research meetings (preparing an agenda and compiling minutes for each meeting), developed and refined research protocols, and wrote quarterly and annual reports that were modified by the PI for submission of required technical reports.

Students gained experience in conducting a hiring process. Each year, 3 students served on a hiring committee for selecting new students. The hiring committee students reviewed applications, recommended a short list for interviews, interviewed applicants from the short list, ranked applicants that were interviewed.

Students gained critical mentoring and leadership experience by participating on a sub-award titled “4-H Career Readiness Pathways”. That project engaged 3 high school students from underrepresented groups in HMB Lab research projects during the 2017 summer. Our students led recruitment, training, mentoring, and assessment activities with the high school students and led outreach efforts at 2 local elementary schools and 1 local high school.

How were the results disseminated to communities of interest?

Results have been disseminated with 1 published journal paper, 18 conference papers, and 6 M.S. theses (4 published / 2 in publishing queue) on Cal Poly’s Digital Commons public repository. See the Bibliography in the Appendix for more details. Additionally, 1 journal paper is in review and 2 journal papers are in preparation (the 2 journal papers in preparation are represented by the Technical Reports in the Appendix).

3 local high school students were funded by a National 4-H Council sub-award to work in the HMB lab during the 2017 summer. The primary focus of this effort was to increase diversity among underrepresented groups in engineering. The high school students assisted HMB staff with several projects, including this project. Upon completion of these summer internships, the high school students presented HMB lab activities, including this project’s activities, at 2 local elementary schools, 1 local high school, and 1 local youth 4-H club.

What do you plan to do during the next reporting period to accomplish the goals?

Nothing to report.

4. IMPACT

What was the impact on the development of the principal discipline(s) of the project?

Novel and/or innovative aspects of the project included the following: 1) reporting select knee kinematic and kinetic parameters of transtibial amputees in cycling and elliptical training exercises, which are two exercises commonly used for rehabilitation and lifelong fitness sustainment; 2) calculating and reporting 3-dimensional knee angles for transtibial amputees in gait, cycling, and elliptical training that were corrected for soft tissue artifact errors; and 3) providing data for evidence-based guidelines for rehabilitation and fitness sustainment exercises for transtibial amputees that minimize intact knee injury and OA.

To date, we presented 18 conference papers but only published 1 journal paper. The journal paper was published in Plos One, which is the journal recommended by the Veterans Administration as the new venue for veteran-centric rehabilitation research publications, replacing the Journal of Rehabilitation & Research Development. As mentioned above, we have

2 journal papers in preparation (represented by the Technical Reports in the Appendix) and completion of our journal publishing aims will ensure that our results will be disseminated broadly. We suggest that the results ultimately published in all of our journal papers lead to an improved understanding of interventions aimed at improving biomechanics, reducing injury, and, ultimately, increasing QoL among both military and civilian transtibial amputees.

What was the impact on other disciplines?

The impact on other disciplines includes the following. 1) The results with control (i.e. non-amputee) participants may also be used to provide evidence-based guidelines for exercises for other populations that are at high risk for knee OA; e.g. overweight and obese individuals. 2) The results suggest that OpenSim is an appropriate analytical tool to use for estimating joint loads in exercises (such as cycling and elliptical) where external loads apart from ground reaction forces are applied to the human body (e.g. by a seat, handlebars). 3) Use and results of novel analytical techniques for calculating knee angles corrected for soft tissue artifact may benefit the broader biomechanical research community. 4) New knowledge has been generated related to knee and cartilage tissue loading in cycling and elliptical exercises.

What was the impact on technology transfer?

Nothing to report.

What was the impact on society beyond science and technology?

We suggest that the primary impact is evidence for the recommendation that non-weight bearing exercises (including cycling) be emphasized for rehabilitation or lifelong fitness sustainment exercise for populations that are at high risk for knee and hip OA.

5. CHANGES/PROBLEMS

Changes in approach and reasons for change / Actual or anticipated problems or delays and actions or plans to resolve them

We experienced difficulty, likely due to our rural location, in recruiting amputee subjects that meet all of our inclusion/exclusion criteria. Originally, the inclusion/exclusion criteria were quite severe in an effort to obtain subjects that would be expected to have normal biomechanics in their intact leg while being eligible for MRI tests. Further, as we suggested in our original proposal, development of the knee FEA models was very challenging. In our original proposal, we clearly stated that it would be difficult to develop knee FEA models for all 20 subjects and thus proposed an alternate plan of developing knee FEA models for only 6 subjects.

To deal with those problems, the following actions were undertaken. 1) We received approval to revise our Statement of Work by restricting our targeted number of MRI models to 6. 2) We obtained Cal Poly's IRB approval to modify our IRB protocol (see below for more details). 3) We requested and received a no-cost extension for 1 year.

Regardless, as noted above, many of our FE model simulations did not successfully converge at the targeted kinetic and kinematic levels (see Section 3 for more details).

Changes that had a significant impact on expenditures

During 2017 we submitted, and received approval for, a budget modification. The greatest modification re-budgeted funds in order to purchase needed equipment, materials, and supplies

that improved our motion analysis capabilities and, consequently, improved the quality and accuracy of our results.

Significant changes in use or care of human subjects

Please see above for the rationale for the following changes. 1) We restricted our targeted number of MRI models to 6; thus, we modified our IRB protocol (and obtained Cal Poly IRB approval) so that the inclusion/exclusion criteria for MRI scans was only applicable for subjects who received a follow-up MRI (e.g. implanted pacemakers and hardware are complications that have excluded subjects from the study solely due to MRI safety). 2) We obtained Cal Poly's IRB approval to increase the maximum age of subjects from 40 to 50 years. This change was not considered to increase risk to human subjects so only Cal Poly IRB approval was needed. 3) We obtained Cal Poly's IRB approval to increase the maximum years since amputation surgery from 10 to 15 years. This change was not considered to increase risk to human subjects so only Cal Poly IRB approval was needed.

6. PRODUCTS

Publications, conference papers, and presentations

Journal publications.

Orekhov, G., Robinson, A. M., Hazelwood, S. J., & Klisch, S. M. (2019). Knee joint biomechanics in transtibial amputees in gait, cycling, and elliptical training. *PLoS ONE*, 14(12), e0226060. <https://doi.org/10.1371/journal.pone.0226060>.

Books or other non-periodical, one-time publications.

Fernandez, R., Biglari, S., Orekhov, G., Robinson, A. M., Hazelwood, S. J., & Klisch, S. M. (2020). Knee and hip joint contact compressive forces in transtibial amputees for gait, cycling, and elliptical training. *Technical Report, HMB Lab, California Polytechnic State University, San Luis Obispo, CA, USA*.

Tucker, S. L., Hazelwood, S. J., & Klisch, S. M. (2020). Knee angles corrected for soft tissue artifact in transtibial amputee during gait. *Technical Report, HMB Lab, California Polytechnic State University, San Luis Obispo, CA, USA*.

Other publications, conference papers, and presentations.

Fernandez, R., Biglari, S., O'Hara, C. R., Hazelwood, S. J., & Klisch, S. M. (2019). Knee joint contact compressive force of transtibial amputee and control participants in gait, cycling, and elliptical. *Southwest American College of Sports Medicine Regional Chapter Annual Meeting*.

Gutierrez-Franco, J. D., Skaro, J. M., Hazelwood, S. J., & Klisch, S. M. (2017). Knee biomechanics during cycling are similar for normal weight and obese subjects. *Summer Biomechanics, Bioengineering & Biotransport Conference*.

Heyde, E. A., Orekhov, G., Robinson, A. M., Hazelwood, S. J., & Klisch, S. M. (2018). Hip and knee forces for transtibial amputees in gait and cycling. *Annual Meeting of the American Society of Biomechanics*.

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Biomechanics.

- Mavrommati, K., Pottinger, M. V., Schueckler, O. J., Hazelwood, S. J., & Klisch, S. M. (2017). Inverse dynamic analysis of knee contact loads for ACL reconstruction patients during gait and cycling. *Annual Meeting of the Biomedical Engineering Society*.
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- Rumery, M. G. (2018). *Subject-specific finite element predictions of knee cartilage pressure and investigation of cartilage material models*. M.S.Thesis, California Polytechnic State University, San Luis Obispo, CA.
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Wash, B. R., Self, B. P., Hazelwood, S. J., & Klisch, S. M. (2019). Verification of Opensim pelvic residuals in seated cycling with handlebar and seatpost load cells. *Annual Meeting of the American Society of Biomechanics*.

Website(s) or other Internet site(s)

For the published journal paper (Orekhov et al., 2019), all data underlying the findings are available from the FigShare database (URL: <https://figshare.com/s/51a35f4f6bf57feca059>).

Technologies or techniques

Nothing to report (see “other products” below).

Inventions, patent applications, and/or licenses

Nothing to report.

Other Products

- 1) Custom TCPE (triangular Cosserat point element) method software (MatLab) and protocol to reduce soft tissue artifact errors.
- 2) Custom PS (Procrustes solution) method software (MatLab) to reduce soft tissue artifact errors.
- 3) Custom PCA (principal component analysis) software (MatLab) to reduce crosstalk errors.
- 4) Custom time sync code (MatLab) to time sync EMG and motion capture data.
- 5) Custom code (MatLab) to combine 2, 6-axis load cell data into a statically equivalent resultant needed for applying a point load on a participant's foot in inverse dynamic analyses.
- 6) MATLAB script that de-identifies MRI scans to preserve patient privacy.
- 7) MATLAB script utilizing GIBBON to turn .stl surface meshes into tetrahedral computational meshes and write FEBio input files.

7. PARTICIPANTS & OTHER COLLABORATING ORGANIZATIONS

What individuals have worked on the project?

Name: Stephen Klisch

Project Role: PI

Nearest person month worked: 12

Contribution to Project: Dr. Klisch managed the project.

Funding Support: Also funded by the Donald E. Bently Center (provided matching release time to work on this research project).

Name: Scott Hazelwood

<i>Project Role:</i>	<i>co-PI</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Dr. Hazelwood managed statistical analysis and finite element aims.</i>
<i>Name:</i>	<i>Brian Self</i>
<i>Project Role:</i>	<i>co-PI</i>
<i>Nearest person month worked:</i>	<i>1</i>
<i>Contribution to Project:</i>	<i>Dr. Self managed aims to instrument bicycle with handlebar and seat post load cells.</i>
<i>Name:</i>	<i>Hemanth Porumamilla</i>
<i>Project Role:</i>	<i>Faculty investigator</i>
<i>Nearest person month worked:</i>	<i>1</i>
<i>Contribution to Project:</i>	<i>Dr. Porumamilla managed aims to instrument bicycle with pedal load cells.</i>
<i>Name:</i>	<i>Larry Coolidge</i>
<i>Project Role:</i>	<i>Staff – tech support</i>
<i>Nearest person month worked:</i>	<i>1</i>
<i>Contribution to Project:</i>	<i>Mr. Coolidge provide technical support for all computer-related equipment.</i>
<i>Name:</i>	<i>Greg Orekhov</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>8</i>
<i>Contribution to Project:</i>	<i>Led experiments and data analysis of ground/pedal reaction forces, knee angles, and knee joint loads.</i>
<i>Name:</i>	<i>Samuel Tucker</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>8</i>
<i>Contribution to Project:</i>	<i>Led experiments and knee angle data analysis using PRB method to reduce soft tissue artifact-induced errors.</i>
<i>Name:</i>	<i>Brad Wash</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>5</i>
<i>Contribution to Project:</i>	<i>Led design and implementation of stationary bike instrumentation. Responsible for comparison of experimentally determined handlebar and seat loads to pelvic residual generated in OpenSim.</i>
<i>Name:</i>	<i>Jordan Skaro</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>5</i>

<i>Contribution to Project:</i>	<i>Led experiments and knee angle data analysis to reduce crosstalk errors.</i>
<i>Name:</i>	<i>Rey Fernandez</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Led OpenSim ID analyses of knee and hip joint contact loads.</i>
<i>Name:</i>	<i>Shaida Biglari</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Assisted with OpenSim ID analyses of knee and hip joint contact loads.</i>
<i>Name:</i>	<i>Megan Pottinger</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Assisted with experiments and data analysis to determine knee joint loads.</i>
<i>Name:</i>	<i>Katherine Mavrommati</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Assisted with experiments and data analysis to determine knee joint loads.</i>
<i>Name:</i>	<i>Elizabeth Heyde</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Assisted with experiments and ID analyses.</i>
<i>Name:</i>	<i>Scott Weinhardt</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Assisted with analyses to reduce soft tissue artifact and crosstalk errors in knee angle data for control and amputee participants.</i>
<i>Name:</i>	<i>Alejandro Gonzalez-Smith</i>
<i>Project Role:</i>	<i>Student researcher</i>
<i>Nearest person month worked:</i>	<i>4</i>
<i>Contribution to Project:</i>	<i>Assisted with experiments and knee angle data analysis using PRB method to reduce soft tissue artifact-induced errors.</i>

Name: Nina Yadlowsky
Project Role: Student researcher
Nearest person month worked: 3
Contribution to Project: Assisted with experiments and knee angle data analysis using PRB method to reduce soft tissue artifact-induced errors.

Name: Emily Vassilev
Project Role: Student researcher
Nearest person month worked: 1
Contribution to Project: Assisted with experiments and data analysis.

Name: Sam Porter
Project Role: Student researcher
Nearest person month worked: 1
Contribution to Project: Assisted with experiments and data analysis.

Name: Ryan Sax
Project Role: Student researcher
Nearest person month worked: 1
Contribution to Project: Assisted with experiments and data analysis.

Name: Emily Hubbard
Project Role: Student researcher
Nearest person month worked: 1
Contribution to Project: Assisted with experiments and data analysis.

Name: Michael Rumery
Project Role: Student researcher
Nearest person month worked: 11
Contribution to Project: Developed finite element models and analyses to determine cartilage tissue loading.

Name: Greg Lane
Project Role: Student researcher
Nearest person month worked: 8
Contribution to Project: Developed finite element models and analyses to determine cartilage tissue loading.

Name: Jonathon Stearns
Project Role: Student researcher
Nearest person month worked: 5
Contribution to Project: Conducted finite element analyses of control and amputee participants in gait, cycling, and elliptical training.

Has there been a change in the active other support of the PD/PI(s) or senior/key personnel since the last reporting period?

Nothing to report.

What other organizations were involved as partners?

Nothing to report.

8. SPECIAL REPORTING REQUIREMENTS

Nothing to report.

9. APPENDIX

Bibliography

- Fernandez, R., Biglari, S., O'Hara, C. R., Hazelwood, S. J., & Klisch, S. M. (2019). Knee joint contact compressive force of transtibial amputee and control participants in gait, cycling, and elliptical. *Southwest American College of Sports Medicine Regional Chapter Annual Meeting*.
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The journal paper corresponding to reference (Orekhov et al., 2019) (pages 20-35), the internal technical report corresponding to (Fernandez et al., 2020) (pages 36-38), the internal technical report corresponding to (Tucker et al., 2020) (pages 39-40), and the MS thesis (Stearns, 2020) (pages 41-139) (in that order) are included in the remaining pages of this Appendix.

RESEARCH ARTICLE

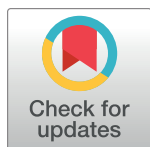
Knee joint biomechanics in transtibial amputees in gait, cycling, and elliptical training

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1 Mechanical Engineering Department, California Polytechnic State University, San Luis Obispo, CA, United States of America, **2** Hanger Clinic, San Luis Obispo, CA, United States of America, **3** Biomedical Engineering Department, California Polytechnic State University, San Luis Obispo, CA, United States of America

 These authors contributed equally to this work.

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OPEN ACCESS

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Data Availability Statement: All data underlying the findings are available from the FigShare database (URL: <https://figshare.com/s/51a35f4f6bf57feca059>).

Funding: The U.S. Army Medical Research Acquisition Activity, 820 Chandler Street, Fort Detrick MD 21702-5014 is the awarding and administering acquisition office. This work was supported by the Defense Health Program, through the Department of Defense Broad Agency Announcement for Extramural Medical Research Program Number W81XWH-BAA-14-1 under

Abstract

Transtibial amputees may experience decreased quality of life due to increased risk of knee joint osteoarthritis (OA). No prior studies have compared knee joint biomechanics for the same group of transtibial amputees in gait, cycling, and elliptical training. Thus, the goal of this study was to identify preferred exercises for transtibial amputees in the context of reducing risk of knee OA. The hypotheses were: 1) knee biomechanics would differ due to participant status (amputee, control), exercise, and leg type (intact, residual) and 2) gait kinematic parameters would differ due to participant status and leg type. Ten unilateral transtibial amputee and ten control participants performed exercises while kinematic and kinetic data were collected. Two-factor repeated measures analysis of variance with post-hoc Tukey tests and non-parametric equivalents were performed to determine significance. Maximum knee compressive force, extension torque, and abduction torque were lowest in cycling and highest in gait regardless of participant type. Amputee maximum knee extension torque was higher in the intact vs. residual knee in gait. Amputee maximum knee flexion angle was higher in the residual vs. intact knee in gait and elliptical. Gait midstance knee flexion angle timing was asymmetrical for amputees and knee angle was lower in the amputee residual vs. control non-dominant knees. The results suggest that cycling, and likely other non-weight bearing exercises, may be preferred exercises for amputees due to significant reductions in biomechanical asymmetries and joint loads.

Introduction

Transtibial amputees may experience decreased quality of life due to increased risk of knee joint osteoarthritis (OA) [1], a degenerative disease of bone and cartilage tissues that often leads to debilitating joint stiffness and pain. Abnormal biomechanics during daily activities, including gait, is well accepted to be a risk factor for knee joint OA [2–5], and is likely related to the high prevalence of joint pain and OA among military and civilian unilateral transtibial and transfemoral amputees [1,6–10].

award no. W81XWH-16-1-0051 (SK) and by the Donald E. Bentley Center (<https://me.calpoly.edu/bently-center>) (SK). Opinions, interpretations, conclusions and recommendations are those of the author and are not necessarily endorsed by the Department of Defense. The funders had no role in study design, data collection and analysis, decision to publish, or preparation of the manuscript.

Competing interests: A. Matt Robinson declares an author commercial affiliation to Hanger Clinic. Mr. Robinson is a licensed Hanger Clinic prosthetist and contributed to study design and implementation. This does not alter our adherence to PLOS ONE policies on sharing data and materials.

Transtibial amputees are more likely to develop OA in the intact knee joint than the residual (i.e. amputated) joint [1,11,12], which may be caused by preferring the intact leg as evidenced by abnormal gait biomechanics including asymmetric ground reaction forces, muscle activation patterns, and knee joint kinetics (i.e. forces and torques) between the intact and residual limbs [13–21]. Studies have shown that transtibial amputees, as compared to non-amputee controls, have increased intact knee joint forces and torques [22] and increased asymmetry in internal knee abduction torque [23,24]. Knee OA is more commonly seen in the medial compartment and, generally, high internal abduction torque increases medial compartment loading [25]. Although there is contention about the relevance of knee extension torque in OA risk [26], an *in vivo* study showed that extension torque is a significant contributor to medial contact force [27].

While prosthetic limb design has advanced in recent years, with energy storage and release (ESAR) prostheses representing the state-of-the-art in passive-elastic devices, such prostheses do not completely restore natural biomechanics of the lower limb during gait [28–30]. Studies comparing powered ankle prostheses to passive devices have shown that increasing net positive prosthetic ankle work can decrease metabolic cost of transport, increase walking speed, and potentially reduce some gait abnormalities [31–35], but the number of studies of powered prostheses is relatively low and these studies are limited to gait only. This study focuses on ESAR prostheses due to the prevalence of passive-elastic devices amongst the amputee population.

Exercise is recommended for amputees for rehabilitation and lifelong fitness sustainment [36] but there are only a few biomechanical studies for transtibial amputees in exercises other than gait. Although guidelines have been proposed for prosthetic use among transtibial amputees during cycling [37], only a few studies have addressed intact knee joint loading during cycling for this population [8,38]. In a prior cycling biomechanics study, significantly higher pedal force and work asymmetries existed between intact and residual limbs for transtibial amputees [38] and such asymmetries depended on cycling intensity and prosthetic foot stiffness [8]. Although elliptical training has been recommended during rehabilitation for this population [39], there do not appear to be any biomechanical studies of knee joint loading during elliptical training for transtibial amputees. Previous studies with non-amputee populations have shown that elliptical training, compared to walking, produces similar kinematic and kinetic patterns [40,41] and reduces knee load impulses [42]. Elliptical training is recommended alongside cycling and swimming as non-weightbearing exercises when other exercises are challenging in persons with joint OA [43].

The goal of this study was to conduct motion analysis experiments and compare knee biomechanics in gait, cycling, and elliptical training in order to identify appropriate exercises for transtibial amputees whom are at high risk for knee OA. The hypotheses were: 1) knee biomechanics would differ due to participant status (amputee, control), exercise, and leg type (intact, residual) and 2) gait kinematic parameters would differ due to participant status and leg type.

Methods

All protocols were approved by Cal Poly's Institutional Review Board (IRB) and the U.S. Army Medical Research and Materiel Command (USAMRMC) Office of Research Protections (ORP) Human Research Protections Office (HRPO) (HRPO Log Number A-19263) and were designed to minimize risk to human subjects. Written consent was obtained prior to all experiments.

Participant recruitment

Ten unilateral transtibial amputees (aged 18–45, body mass index [BMI] 22.3–29.6, 1.5–12.4 years post-operation, 7 males, 3 females) and ten control participants (aged 20–26, BMI 19.1–

Table 1. Transtibial amputee participant characteristics.

Participant	Age	Mass [kg]	Height [m]	BMI	Intact Leg	Gender	Years Since Amputation	Co-Morbidities
1	32	74.8	1.82	22.6	L	M	11.9	Healed ACL tear
2	31	83.9	1.69	29.4	R	M	2.8	-
3	45	80.7	1.81	24.6	R	M	6.2	Charcot foot
4	32	80.6	1.78	25.4	R	M	12.4	Screws in ankle
5	32	72.4	1.80	22.3	R	M	8.1	-
6	34	92.2	1.77	29.6	L	F	10.6	-
7	29	58.4	1.54	24.7	L	F	4.2	-
8	37	82.9	1.82	25.0	L	M	8.2	Hip implant, rod in femur
9	18	76.8	1.73	25.5	R	F	1.5	-
10	32	75.0	1.81	22.8	R	M	6.8	-
Average (SD)	32.2 (6.7)	77.8 (8.9)	1.76 (0.09)	25.2 (2.6)	-	-	7.3 (3.7)	-

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27.9, 8 males, 2 females) participated. Exclusion criteria included history of cardiovascular disease, respiratory disease, or any other metabolic disease/complication; substantial weight loss or gain over the previous six months; history of major psychiatric illness, drug abuse, or unsafe dieting practices; major medical conditions that prohibit physical activity; and pregnant women or women expecting or trying to be pregnant. After screening, eligible participants were invited to the motion analysis lab where informed consent was obtained, information forms were filled out, and amputee participants were fitted with an Energy Storage and Release (ESAR) prosthesis (Vari-Flex[®], ssur, Reykjavik, Iceland) by a certified prosthetist (AMR). One week of accommodation time was provided to amputee participants that exhibited unfamiliarity with the Vari-Flex[®] as determined by our certified prosthetist (AMR) in qualitative alignment and gait pattern checks. However, most (8/10) amputee participants were tested without accommodation time as they used similar ESAR devices to the Vari-Flex[®] and passed AMR's alignment and gait pattern checks. Amputee participants all wore similar carbon fiber sockets and static pylons with no springs or dampers. Screening and health information forms indicated that all participants were relatively young, healthy, non-obese, free of disease, and had no restrictions on physical activity.

Tables 1 and 2 document critical information for amputee and control participants, respectively. Mass was measured with a scale and height measured with a stadiometer. The dominant

Table 2. Control participant characteristics.

Participant	Age	Mass [kg]	Height [m]	BMI	Dominant Leg	Gender
1	26	86.6	1.79	27.0	R	M
2	23	88.5	1.75	23.0	L	M
3	22	61.2	1.79	19.1	R	M
4	22	79.1	1.82	24.0	R	M
5	20	90.4	1.80	27.9	L	M
6	23	70.3	1.75	22.9	R	M
7	23	56.2	1.63	21.3	R	M
8	21	64.8	1.65	23.9	R	M
9	20	68.1	1.61	26.1	R	F
10	21	68.7	1.72	23.4	R	F
Average (SD)	22.1 (1.8)	73.4 (12.0)	1.73 (0.08)	24.4 (3.1)	-	-

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leg for amputee participants was defined as the intact leg. The dominant leg for control participants was defined as the strongest or preferred leg and was self-reported. Age, gender, years since amputation, and co-morbidities for the intact leg were self-reported.

Experiments

Twelve near-infrared digital cameras (6 Owl, 3 Osprey, 2 Kestrel, 1 Eagle) (Motion Analysis Corp., Santa Rosa, CA, USA) were used capture the motion of reflective markers. Participants wore tight compression clothing that exposed as much skin as the participant was comfortable with. A modified Helen Hayes marker set was used with 32 reflective markers placed at the following anatomical landmarks: crown of the head, acromion processes, 7th cervical spine, sternum, greater trochanters, anterior superior iliac spines, tops of the iliac crests, sacrum, halfway along the long axis of each thigh, lateral and medial knee condyles, fibular heads, tibial tuberosities, halfway along the long axis of each shank, lateral and medial malleoli, Achilles tendon insertion sites, and between the 2nd and 3rd metatarsals of both feet. When possible, markers were placed directly on skin but pelvic markers were placed on compression clothing. Prosthesis-side markers were aligned with intact-side markers as closely as possible when anatomical landmarks were inaccessible or absent.

Participants were asked to stand stationary in a static pose with feet at shoulder width and arms bent at the elbows. Static pose captures were used to begin the marker identification process and defined initial joint angles for later analyses. Top head and medial knee and ankle markers were removed after static captures. During gait experiments, four ground force plates (Accugait, AMTI, Watertown, MA, USA) were used to capture ground reaction forces in the anterior, lateral, and vertical directions; the free reaction torque about the vertical axis; and the center of pressure (COP) of the reaction force vector on the surface of the force platform (Fig 1). Three gait trials were collected for each participant at self-selected speeds in each direction.

During cycling and elliptical experiments, the pedals of a stationary bicycle (Lifecycle GX, Life Fitness, Schiller Park, IL, USA) and elliptical trainer (XE795, Spirit Fitness, Jonesboro, AR, USA) were instrumented with 6-axis load cells (AD2.5D, AMTI) to capture pedal reaction forces, free torque, and COP (Fig 1). Three trials for cycling and elliptical were collected at 70

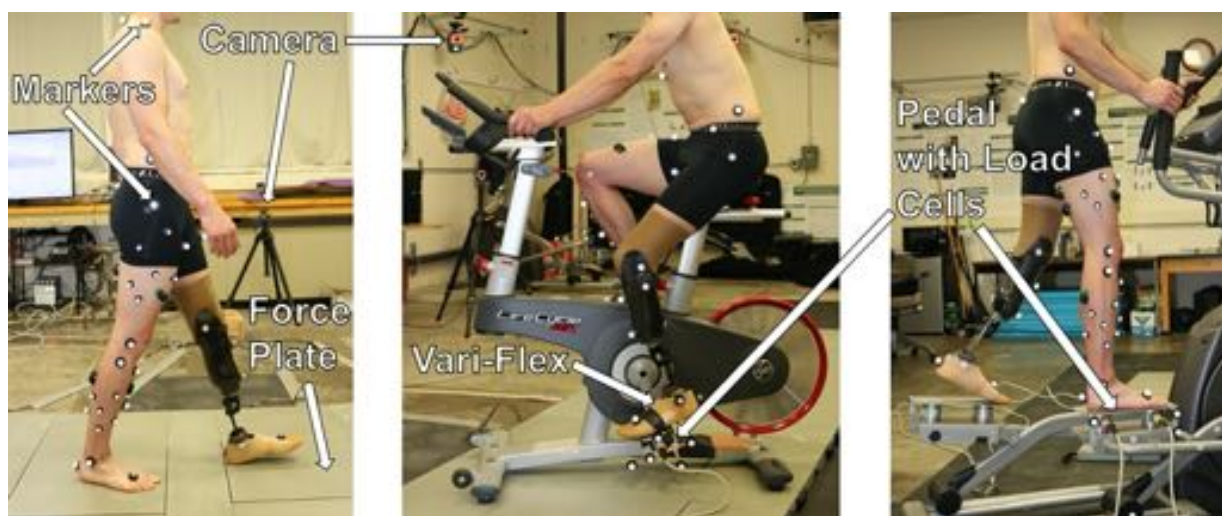


Fig 1. Setup for gait, cycling, and elliptical training motion analysis experiments. Digital cameras, ground force plates, reflective markers, the Vari-Flex prosthesis with foot cover, and instrumented exercise machine pedals are indicated.

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revolutions per minute (RPM, range 68–72 RPM) as measured by the corresponding machine at moderate resistance settings (level 10 of 20). Once the participant reached the target speed, data for 30 seconds were collected to capture several machine cycles. Cortex software (Version 7.01, Motion Analysis) was used to interface all equipment, perform post-processing of marker data, and calculate kinematics and kinetics.

Data processing

One full gait cycle was defined by consecutive heel strikes of the same leg (0% = 1st heel strike, 100% = 2nd heel strike). One cycling cycle was defined by one full crank rotation starting at top dead center (TDC; 0% = 1st pass through TDC, 100% = 2nd pass through TDC) [8]. One elliptical cycle was defined by one full crank rotation starting at the most anterior pedal position (APP; 0% = 1st pass through APP, 100% = 2nd pass through APP) [42].

The sacrum marker velocity in the direction of motion was stored and used to calculate average walking speed for each participant. The time in seconds at which cycling and elliptical pedals passed through their respective start and end of cycle were stored and used to calculate actual RPM. Cycles with RPM greater than 72 or less than 68 were eliminated from the pool of samples as kinetics vary with machine power [8]. Machine cycles were selected randomly without replacement (to avoid selecting the same cycle again) before normalizing and averaging data.

Static and dynamic captures were processed in Cortex to define body segments used to calculate joint kinematics and kinetics. For cycling and elliptical trials, additional markers defined pedal segments which were necessary to track load cells relative to a global coordinate system and transform measured forces and torques to the foot coordinate system (Fig 2). For the elliptical, two load cells were attached to each pedal and their data were combined into an equivalent force-couple system using standard equations [44,45] in order to properly apply reaction forces to the foot (Fig 3).

Kinematic and kinetic data were captured in Cortex at 150 Hz. Marker trajectories were filtered in Cortex using a two-pass, 4th order, zero phase shift Butterworth filter with a cutoff frequency of 6 Hz and reaction forces were filtered using the same filter with a 10 Hz cutoff frequency. Cutoff frequency for reaction forces was calculated using an optimal method [46]. Missing marker data were interpolating using a cubic spline in Cortex.

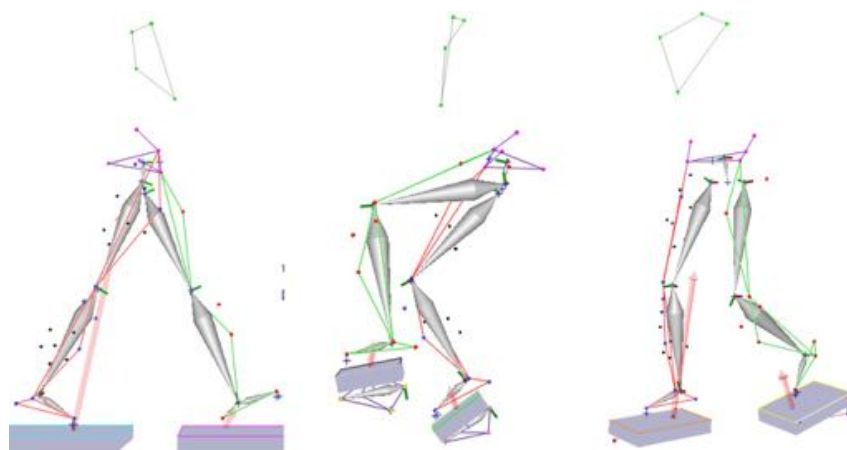


Fig 2. Gait, cycling, and elliptical dynamic captures in Cortex. Force plates and load cells tracking pedal mass segments are visible. Red vectors indicate magnitude and direction of reaction forces.

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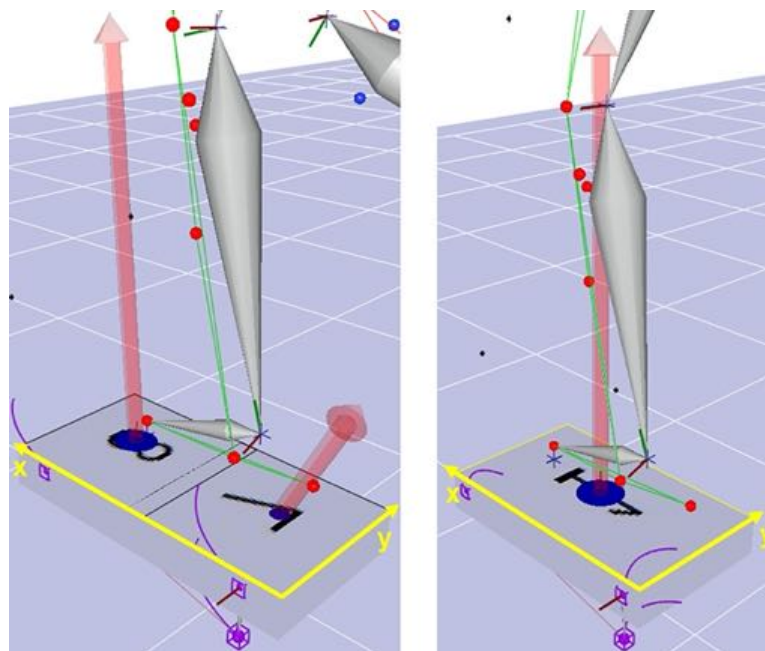


Fig 3. Elliptical force platforms in Cortex before (left) and after (right) combining load cell data into a resultant force-couple system.

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Kinetics were calculated in Cortex using a bottom-up approach. The Cortex package Kin-Tools RT used lower-body kinematics, standard shank and foot segment inertial properties [47], and measured reaction forces to calculate knee joint forces and torques via Euler's equations [48]. The knee joint angles, resultant forces, and resultant torques for the knee were resolved into a floating axis joint coordinate system [49] (Fig 4). Knee flexion angle; lateral, anterior, and compressive force; and extension and abduction torque were defined to be positive. Reported torques follow the internal convention; that is, internal joint torques caused by muscle and joint contact forces oppose external torques due to ground or pedal reaction forces [50].

Analyses of kinematic and kinetic data results were performed using a custom script in MATLAB (MathWorks, Natick, MA, USA). All dynamic trials were interpolated to 101 data points from 0% to 100% cycle. Three trials for each participant in each exercise for each of the intact/dominant and residual/non-dominant legs were averaged. Knee kinematics and kinetics were then averaged across ten participants in each participant group.

Maximum knee flexion angles in gait, cycling, and elliptical for the intact/dominant and residual/non-dominant legs were normalized for each participant group by subtracting calculated knee joint flexion angles of the static pose from the dynamic angles measured during each trial [51]. Also, midstance and swing knee flexion angle maximums and corresponding timing in percent cycle in gait for amputee and control groups were analyzed. Forces were normalized by body weight and torques were normalized by body weight times height [52].

Statistics

Analyzed data were tested for normality using Shapiro-Wilks tests and for equal variance using F-tests. Maximum knee compression force, extension torque, abduction torque, and maximum swing knee flexion angle in gait were found to be non-normal distributions and

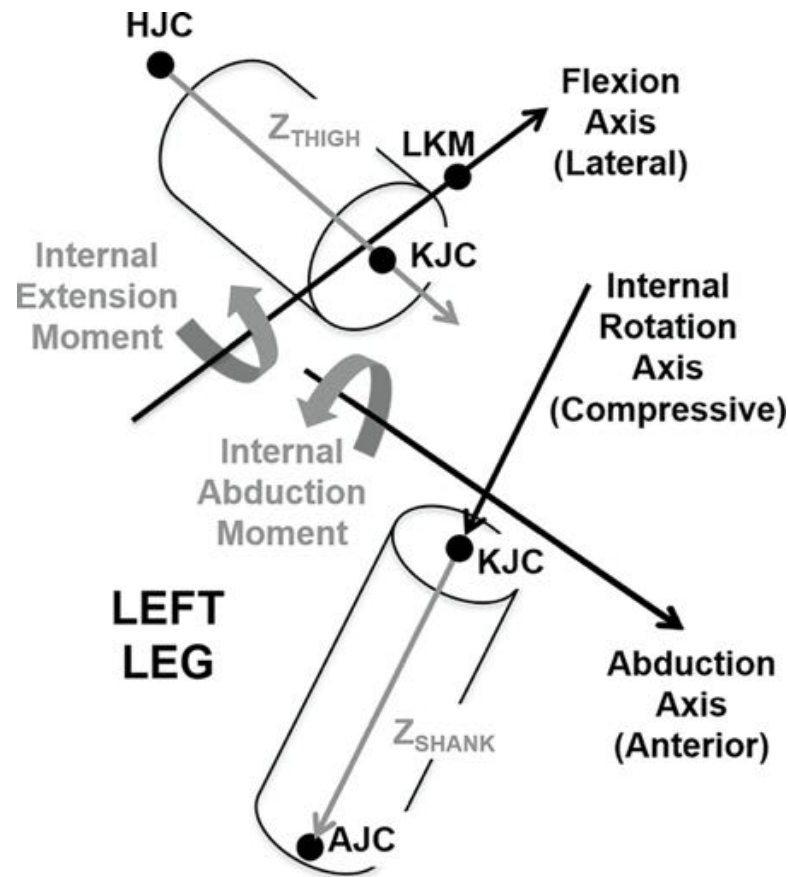


Fig 4. Floating axis knee joint coordinate system. Angle and force axes are indicated in the positive direction in black. Joint torques are defined in the positive orientation in gray. HJC is hip joint center, KJC is knee joint center, and AJC is ankle joint center. "Z" vectors indicate the longitudinal axis of the thigh and shank. The knee internal rotation axis is defined as the shank axis. The flexion axis is defined as the axis perpendicular to the thigh axis pointing through the lateral knee marker (LKM). The abduction axis is the cross product of the internal rotation axis and the flexion axis and points anteriorly.

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were analyzed using Kruskal-Wallis tests and post-hoc Dunn tests at 95% confidence with Bonferroni adjustments for multiple comparisons to analyze the differences between participant-leg-exercise groupings. Two-factor repeated measures analysis of variance (ANOVA) with post-hoc Tukey tests at 95% confidence were conducted to analyze the effects of participant and leg type (amputee intact leg, amputee residual leg, control dominant leg, control non-dominant) and exercise (gait, cycling, elliptical) on maximum knee flexion angle and to analyze the effects of participant and leg type (amputee intact, amputee residual, control dominant, control non-dominant) on knee flexion angle peaks and peak times in gait. All statistical analyses were performed in R [53]. A Bonferroni correction of four was applied (on account of four dependent variables) such that $p < 0.0125$ was considered statistically significant.

Results

The average amputee and control participant walking speeds were 1.26 ± 0.17 m/s and 1.29 ± 0.08 m/s, respectively. Average cycling RPMs were 69.45 ± 0.61 and 69.75 ± 0.61 for amputees and controls, respectively. Average elliptical RPMs were 70.32 ± 0.59 and 70.15 ± 0.54 for amputees and controls, respectively. A two-sample t-test showed that the

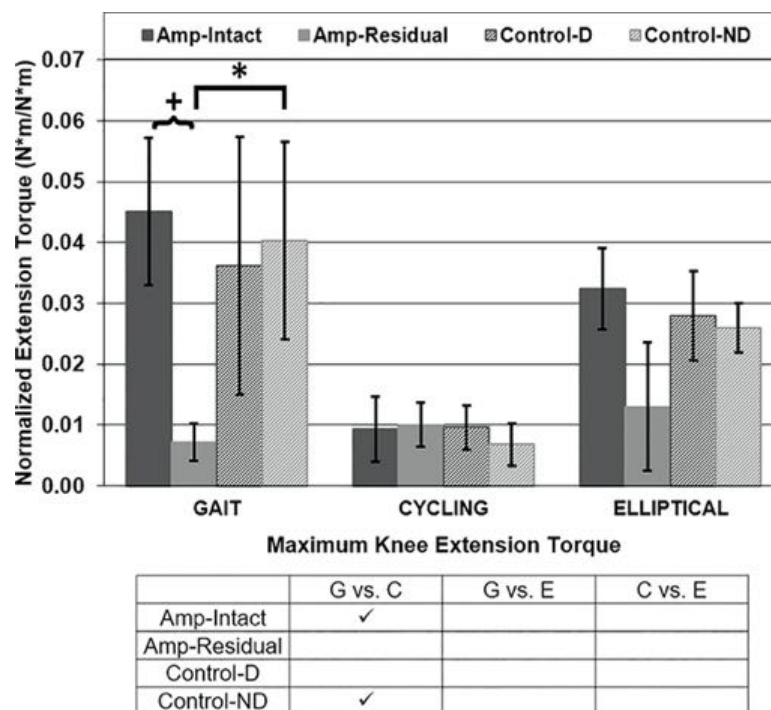


Fig 5. Maximum knee flexion angle [deg]. Mean $\pm$ 1 standard deviation. + = significance across leg type (intact/dominant vs. residual/non-dominant). * = significance across participant type (amputee vs. control). ✓ = significance across exercise type (gait vs. cycling vs. elliptical). $P < 0.0125$ significant.

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walking speeds were not different ($p = 0.566$) between participant groups (t-tests were only run on gait speeds since they were self-selected).

Maximum knee flexion angles were different between exercise types for all participant groups (Fig 5). P-values for maximum knee flexion angles between exercises (gait vs. cycling, gait vs. elliptical, cycling vs. elliptical) for all participant-leg combinations (amputee intact leg [Amp-Intact], amputee residual leg [Amp-Residual], control dominant leg [Con-D], control nondominant leg [Con-ND]) were all < 0.001 . Maximum knee flexion angles were asymmetrical for amputee participants in gait ($p = 0.005$) and elliptical ($p < 0.001$) but not cycling ($p = 0.014$) (Fig 5).

Maximum knee compressive forces varied in gait vs. cycling for all participant types and in cycling vs. elliptical for Amp-Intact legs (Fig 6). P-values for maximum knee compressive force comparisons in gait vs. cycling for all participant types were < 0.001 . Maximum knee compressive forces were not different between participant types or between intact/dominant and residual/non-dominant legs of either participant group (Fig 6).

Maximum knee internal extension torque was different in gait vs. cycling for Amp-Intact and Con-ND legs and in cycling vs. elliptical for Amp-Intact legs (Fig 7). P-values for maximum knee extension torque comparisons in gait vs. cycling were < 0.001 for Amp-Intact and Con-ND legs. Maximum knee extension torque for amputee participants was asymmetrical in gait ($p < 0.001$) (Fig 7). Maximum residual/non-dominant knee extension torque was different between amputees and controls in gait ($p < 0.001$) (Fig 7).

Maximum knee internal abduction torque was different in gait vs. cycling for all participant types (Fig 8). P-values for maximum knee abduction torque comparisons in gait vs. cycling were < 0.001 for all participant-leg combinations. Maximum knee abduction torque was not

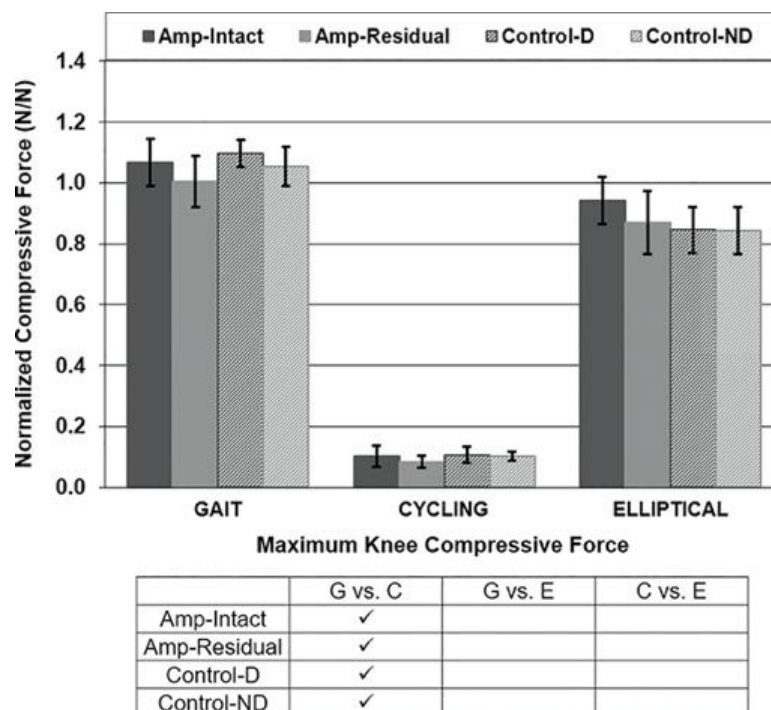


Fig 6. Maximum knee compressive force [N/N]. Mean $\pm$ 1 standard deviation. + = significance across leg type (intact/dominant vs. residual/non-dominant). * = significance across participant type (amputee vs. control). ✓ = significance across exercise type (gait vs. cycling vs. elliptical). $P < 0.0125$ significant.

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different between participant types or between intact/dominant and residual/non-dominant legs of either participant group (Fig 8).

P-value for maximum midstance knee flexion angle in gait was 0.005 for Amp-Residual vs. Con-ND legs (Fig 9). P-values for midstance knee flexion angle timing were <0.001 for both Amp-Intact vs. Amp-Residual legs and Amp-Residual vs. Con-ND legs (Fig 9). P-value for maximum swing knee flexion angle in gait was 0.001 for Amp-Intact vs. Amp-Residual legs (Fig 9). No differences in swing knee flexion angle timing were found.

Discussion

The hypotheses of the current study were: 1) knee biomechanics would differ due to participant status (amputee, control), exercise, and leg type (intact, residual) and 2) gait kinematic parameters would differ due to participant status and leg type. Kinematics and kinetics varied with exercise (all variables), leg type (flexion angle, extension torque), and participant type (extension torque in gait) providing support for the first hypothesis. Gait characteristics varied with leg type (midstance flexion angle timing, swing flexion angle) and participant type (midstance flexion angle, swing flexion angle) providing support for the second hypothesis. Knee joint compressive force, torques, and flexion angles were of similar magnitude to many other studies in both amputee and non-amputee populations in gait [16,17,21,54], cycling [55,56], and elliptical training [41,42,57] (see below for more details).

The kinematic results of the present study have clinical implications regarding rehabilitative and lifelong fitness sustainment exercises that may minimize OA risk in amputees. In the general population, abnormal knee kinematics are linked to the initiation of knee OA [2,58]. Specifically, knee flexion angle and timing are particularly linked to knee OA in the general

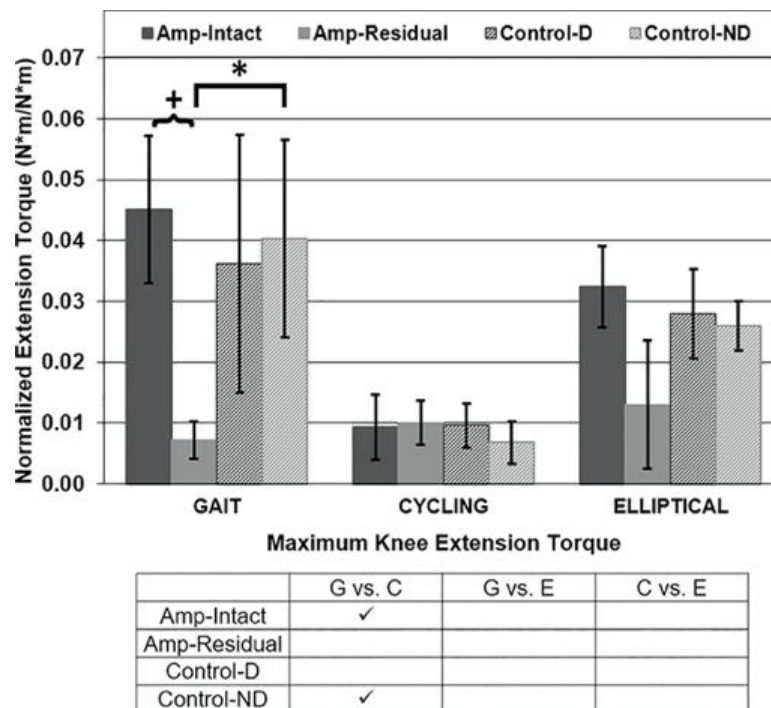


Fig 7. Maximum knee internal extension torque [Nm/Nm]. Mean $\pm$ 1 standard deviation. + = significance across leg type (intact/dominant vs. residual/non-dominant). * = significance across participant type (amputee vs. control). ✓ = significance across exercise type (gait vs. cycling vs. elliptical). $P < 0.0125$ significant.

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population due to evidence that low midstance flexion angle and altered temporal characteristics will cause abnormal knee kinetics [59]. In this study, the amputee intact knee flexion angle was not different from that of control participants during the stance phase in gait. However, maximum residual knee flexion angle was significantly lower in midstance and had significantly delayed timing compared to the intact leg and the control non-dominant leg. Also, swing flexion angle was higher in the residual leg compared to the intact leg. These observed kinematic abnormalities in the residual leg may require compensation which could affect load bearing in the intact knee. For example, the observed reduction in residual knee midstance flexion angle could be evidence of an avoidance strategy (quadriceps-avoidance gait) also seen in anterior cruciate ligament (ACL) deficient populations that are at increased risk for intact knee OA [59]. Compensatory muscle activity in transtibial amputee gait, namely asymmetry in intact vs. residual knee flexor/extensor activity, has been found in previous studies [13,60].

Also, the kinetic results of the present study have similar clinical implications. Knee kinetics such as compressive force and muscle torques can be an indication of OA severity [60,61]; specifically, progression of cartilage degeneration is exacerbated by abnormal kinetics as increased cartilage compressive stress has been shown to be both a symptom of OA and a cause for further degeneration [2,58,61,62]. Abduction torque was found to have a particularly close link to the progression of knee OA and was of higher magnitude in cases of severe medial compartment OA [2,58,60,61,63] and tends to increase with increasing disease severity [26,64,65]. In the current study, no differences of amputee vs. control intact/dominant knee compressive force, extension torque, or abduction torque were found in any of the exercises. However, large asymmetries were present in maximum extension torque for amputees in gait suggesting that muscle coordination and braking/propulsion effort may be altered for the residual leg.

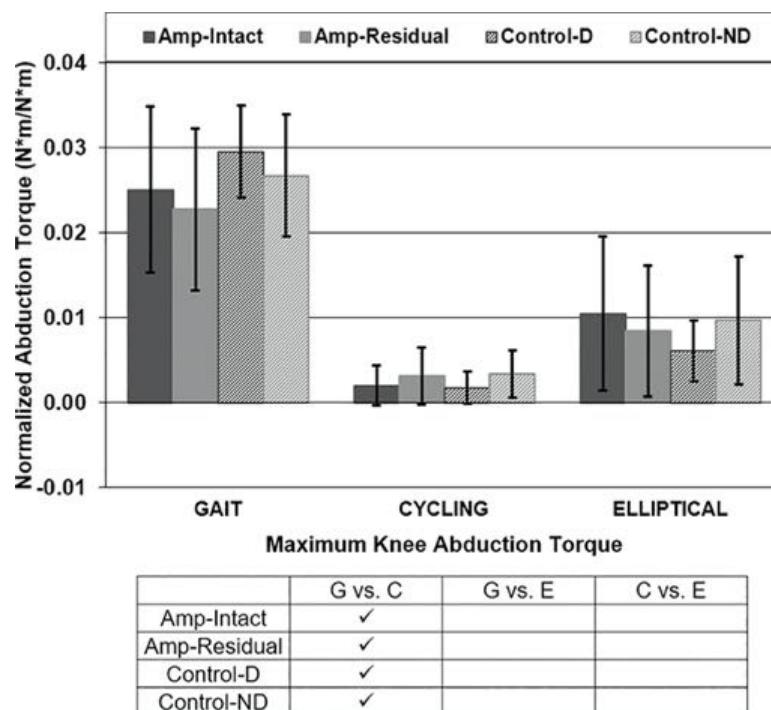


Fig 8. Maximum knee internal abduction torque [Nm/Nm]. Mean $\pm$ 1 standard deviation. + = significance across leg type (intact/dominant vs. residual/non-dominant). * = significance across participant type (amputee vs. control). ✓ = significance across exercise type (gait vs. cycling vs. elliptical). $P < 0.0125$ significant.

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Significantly reduced residual knee extension torque may be a sign that ESAR prostheses do not adequately replace natural biomechanics after amputation and may be causing the intact

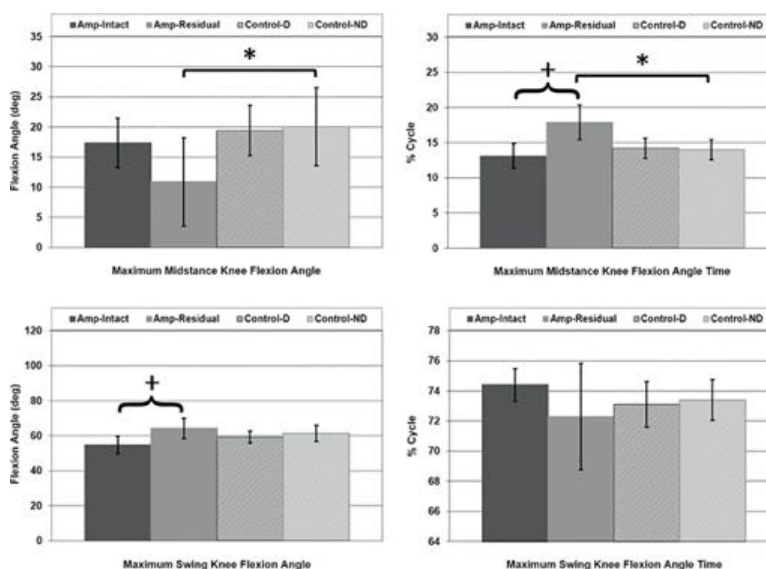


Fig 9. Maximum midstance and swing knee flexion angle [deg] and corresponding times [% Cycle] in gait. Mean $\pm$ 1 standard deviation. Midstance results are in the first row, swing results are in the second row, flexion angle results are in the first column, and flexion angle time results are in the second column. + = significance across leg. * = significance across participant type. $P < 0.0125$ significant.

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leg to compensate. Also, knee kinetics were generally lowest in cycling and highest in gait, suggesting that for populations at high risk for knee OA (such as amputees), cycling and other non-weight bearing exercises may be preferred for rehabilitation and lifelong fitness sustainment. However, this study did not directly link exercise type to injury or OA risk and a long-term study to identify evidence-based exercise guidelines for limiting OA risk is needed.

The results of the current study suggest that cycling may be an appropriate exercise for transtibial amputees whom are at high risk for knee OA. Specifically, the results suggest that exercises that constrain kinematics, such as cycling, are more likely to maintain typical cartilage loading patterns due to a lack of knee flexion angle and extension torque asymmetry as compared to gait and elliptical. The results also suggest that exercises that reduce overall knee joint forces and torques may be preferred for reducing OA risk. Cycling had generally lower magnitudes of resultant knee compressive force, extension torque, and abduction torque compared to elliptical training and gait. Other non-weight bearing exercises such as rowing or non-impact exercises such as stair-stepping may also be preferred over gait, while high-impact activities such as running and sports that involve foot planting (such as soccer) may increase the rate of OA progression due to high knee joint compression forces. Exercises for rehabilitation should, in general, minimize kinematic abnormalities and joint compressive stress to prevent or alleviate abnormal loading of cartilage but the present study did not relate exercise type to injury or OA risk. There is a clear need for a long-term study to associate injury risk with varying exercise protocols to produce specific exercise guidelines for limiting OA risk.

There are several limitations to the current study. One limitation of this work included marker-based errors which are common in motion analysis. Markers may have moved relative to the underlying bones due to soft tissue artifact (STA) and relative motion of compression clothing. STA, caused by marker movement due to skin deformation and displacement, affects the estimation of skeletal system kinematics with the exception of motion about joint flexion-extension axes [66,67]; this limitation was mitigated by only considering knee flexion angles. Markers defining the pelvis were placed on compression clothing; motion of these markers relative to the body would affect the calculation of the hip joint center which may impact knee flexion angle calculations; this limitation was mitigated by proper clothing fit. Another marker-based error, kinematic crosstalk, occurs when the calculated joint coordinate system is misaligned with anatomical axes which significantly influences knee joint abduction and internal rotation angles [68]; hence only knee flexion angles were considered. Another source of marker-based error was marker obstruction or loss during data collection; gaps in data were interpolated using a cubic spline.

A second limitation was inherent issues in any prosthetic assembly for amputee participants. While the same ESAR foot was used for all amputee participants the socket was not modified to enhance user experience and comfort with the prosthesis in exercises that amputee participants were not familiar with (cycling and elliptical). Socket-limb interfaces (such as liners, sleeves, or vacuum-based sockets) can permit motion of the prosthesis relative to the residual limb which can affect joint biomechanics. Future studies should make efforts to measure prosthesis motion relative to the residual limb.

In conclusion, transtibial amputees had significant asymmetry between intact and residual knee flexion angle in gait and elliptical and significantly reduced extension torque in the residual vs. intact knee in gait, whereas no asymmetries were detected for transtibial amputees in cycling. The results suggest that cycling, and likely other non-weight bearing exercises, may be a preferred exercise for limiting OA risk in transtibial amputees due to reduced asymmetry in knee kinematics and reduced knee kinetics as compared to gait values. Also, the results showed that state-of-the-art ESAR prosthetic design may not sufficiently restore amputee biomechanics to normal levels as evidenced by midstance knee flexion angle and peak extension torque

asymmetry in gait and peak knee flexion angle asymmetry in elliptical. Muscle activation patterns could be responsible for altered residual knee flexion angle and extension torque in amputees and future work should involve EMG-driven inverse dynamics to calculate intact knee joint contact forces and muscle contributions. Since this study did not directly relate exercise type to injury or OA risk, there is a need for a long-term study that aims to directly associate injury or OA risk with varying exercise protocols for transtibial amputees and other populations at high risk for knee OA.

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Writing – original draft: Greg Orekhov.

Writing – review & editing: Scott J. Hazelwood, Stephen M. Klisch.

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KNEE AND HIP JOINT CONTACT COMPRESSIVE FORCES IN TRANSTIBIAL AMPUTEES FOR GAIT, CYCLING, AND ELLIPTICAL TRAINING

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INTRODUCTION

Transtibial amputees have an increased risk of contralateral (intact) knee osteoarthritis (OA) [1,2] that may be influenced by abnormal knee and hip joint contact compressive forces (KJCCF, HJCCF). Studies indicate that TTs have higher resultant knee joint forces compared to non-amputees [3]; however, no studies have reported TT knee and hip JCCFs. The long-term goal of this project was to compare knee and hip JCCFs for exercises that may be considered for rehabilitation and lifelong fitness sustainment guidelines that also consider minimizing OA risk. The objective of this study was to utilize biomechanical modeling software to predict KJCCF and HJCCF for gait (G), cycling (C), and elliptical training (E) exercises. The hypotheses of this study were three-fold: knee and hip joint contact compressive forces would differ due to (H1) exercises for the amputee group (amputee gait vs. amputee cycling vs. amputee elliptical) and (H2) exercises for control group (control gait vs. control cycling vs. control elliptical) and (H3) participant status for each same exercise (amputee- vs. control gait, amputee- vs. control cycling, and amputee- vs. control elliptical).

METHODS

All protocols were approved by Cal Poly's Institutional Review Board (IRB) and were designed to minimize risks to participants. Ten unilateral transtibial amputees (aged 32.2 ± 6.7 , BMI 25.2 ± 2.6 , 7 males, 3 females) and ten control participants (aged 22.1 ± 1.8 , BMI 24.4 ± 3.1 , 8 males, 2 females) performed a minimum of three trials for G, C, and E. An enhanced Helen Hayes marker set with 32 markers were used to identify anatomical landmarks. Kinematic parameters were recorded with 12 cameras at 150 Hz using Cortex software (Motion Analysis Corp., Santa Rosa, CA, USA). Kinematic data were processed and filtered (4th

order Butterworth, 6 Hz cutoff frequency) in Cortex. G trials were conducted at self-selected speeds with 4 force plates (AccuGait AMTI, Watertown, MA, USA). C and E trials were performed at a cadence of 70 RPM at moderate resistance (Level 10 of 20) for 30 seconds using a stationary bike (Lifecycle GX, Life Fitness, Schiller Park, IL, USA) and elliptical machine (XE-795, Spirit Fitness, AR, USA) with custom pedals containing 6-axis load cells (AD2.5D, AMTI). Matlab (MathWorks, Natick, MA, USA) was used to export kinematic and kinetic data for OpenSim (Stanford University, Palo Alto, CA, USA).

A OpenSim musculoskeletal model [4] developed for relatively large knee flexion angles (as observed for C and, to a lesser extent, E) was used. OpenSim's scaling tool was utilized to scale the model for each participant. OpenSim's Inverse Kinematics (IK) tool was used to calculate joint kinematics. Those results were then loaded into OpenSim's Residual Reduction Algorithm (RRA) tool to correct the joint kinematics and optimize the scaled model using ground reaction forces data. OpenSim's Static Optimization (SO) tool to calculate muscle forces that were subsequently inputted to the Joint Reaction (JR) tool to calculate KJCCF and HJCCF.

JR results were interpolated to one full gait cycle (0% = 1st heel strike, 100% = 2nd heel strike) or one crank revolution for cycling (0% = 1st top dead center [0 deg], 100% = 2nd top dead center [360 deg.]) and elliptical (0% = 1st pass most anterior position [0 deg], 100% = 2nd pass most anterior position [360 deg.]). Three trials for each exercise were averaged. Only the dominant and intact legs for control and amputee participants, respectively, were considered. KJCCF and HJCCF were normalized by body weight. Outliers were identified

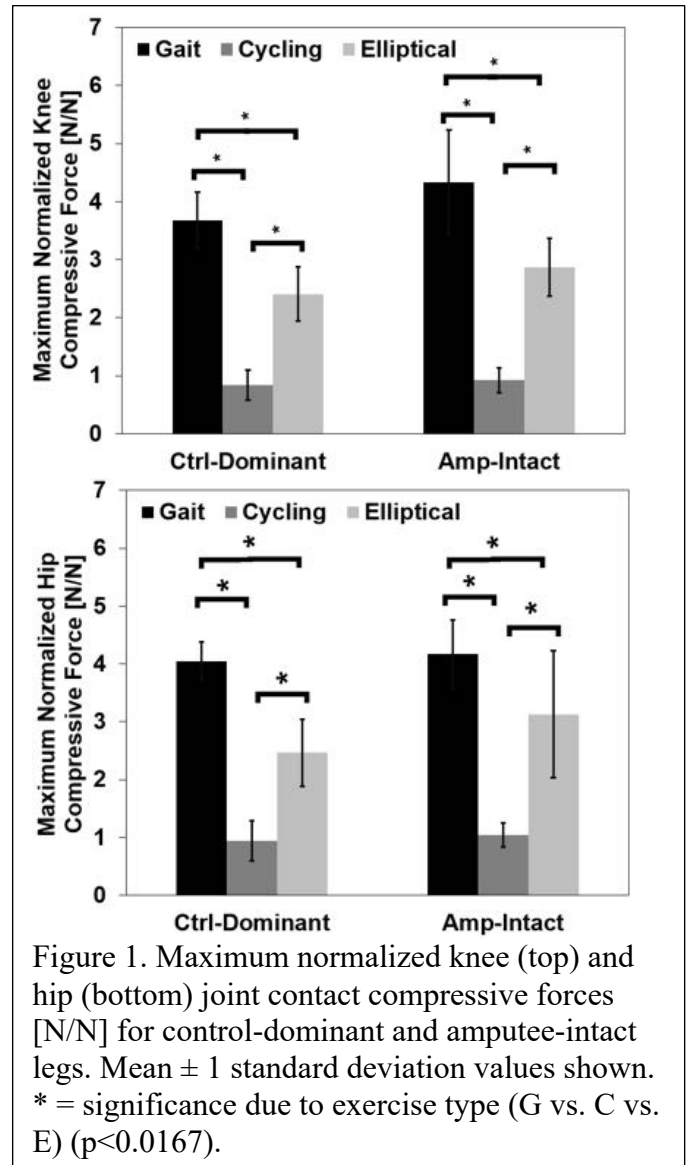
by a two-standard-deviation threshold: KJCCF and HJCCF that were two standard deviations away from the mean were considered outliers and were subsequently removed in statistical analysis.

Significant differences were conducted using a two-factor repeated measures of variance (ANOVA) and post-hoc Tukey tests at 95% confidence. A Bonferroni adjustment of three was applied on the maximum kinetic values due to multiple measures ($p=0.0167$) for each hypothesis.

RESULTS AND DISCUSSION

From the two-standard-deviation test, these n -values were within the outlier threshold: for the control group, maximum KJCCFs and HJCCFs in G ($n=9$, $n=2$), C ($n=9$, $n=10$), and E ($n=9$, $n=8$) were considered; for the amputee group, maximum KJCCFs and HJCCFs in G ($n=10$, $n=10$), C ($n=10$, $n=7$), and E ($n=8$, $n=7$) were analyzed. For each maximum KJCCF and HJCCF, significant differences due to exercise type were found only within a particular population, supporting H1 for amputees and H2 for controls (Fig. 1). Also, KJCCF and HJCCF were consistently highest in G and lowest in C for both control and amputee groups ($G>E$, $E>C$, $G>C$). No differences were found across amputee-intact and control-dominant legs for each exercise (e.g., amputee-intact G vs. control-dominant G). Thus, the results did not support H3 (Table 1).

A clinical implication of the results is that, for TT amputees, cycling should be emphasized over gait and elliptical training for rehabilitation and/or lifelong fitness sustainment exercises that may



minimize OA risk. That conclusion generally agrees with an earlier study [3] which suggested that for TT amputees cycling likely produces normal (i.e. similar to controls) cartilage loading

Table 1. p-values for comparisons of maximum normalized knee and hip joint contact compressive forces [N/N]. * = significance across exercise types within a population (G vs. C vs. E) ($p<0.0167$ significance level). Tabulated p-values ordered as KJCCF p-value / HJCCF p-value. ^a indicates comparison across population type for the same exercise (e.g., G amp-intact vs G ctrl-dominant).

Population	p-Values					
	G vs. C	G vs. E	C vs. E	G	C	E
Amp-Intact	<0.001*/	<0.001*/	<0.001*/			
	<0.001*	0.011*	<0.001*			
Ctrl-Dominant	<0.001*/	0.002*/	<0.001*/			
	<0.001*	<0.001*	<0.001*			
Amp-Ctrl ^a				0.083	0.999	0.462
				1.000	1.000	0.284

patterns due to cycling's status as an exercise that is non-weight bearing and which constrains the feet via a pedal. This study's results also suggest that, for TT amputees, elliptical training should be preferred over gait.

CONCLUSIONS

Novel results of this study were predicting knee and hip joint contact compressive forces for TT amputees for gait, cycling, and elliptical training exercises using OpenSim. The results suggest that non-weight bearing and non-impact activities (such as cycling) are preferred exercises for reducing joint contact loading and, consequently, minimizing OA risk.

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KNEE KINEMATICS FOR THE INTACT LEG OF TRANSTIBIAL AMPUTEES IN GAIT AND CYCLING

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INTRODUCTION

Transtibial (TT) amputees have an increased risk of intact knee osteoarthritis (OA) [1,2] that may be influenced by asymmetries in knee biomechanics [3]. In order to correlate differences in knee kinematics between participant types with physiological and experimental factors (BMI, exercise type, etc.), precise measurement of knee kinematics is critical. Soft tissue artifact (STA) is a considered a leading source of error in motion analysis data collected from skin-based marker systems and introduces errors in calculated knee angles. Previous studies have shown a strong correlation between excess adiposity and lower limb amputation [4]. Due to excessive adipose tissue, STA may introduce considerable errors when analyzing knee angles of TT amputees [5]. Previous studies have reported 3-D knee angles for TT amputees in gait and cycling [6,7] but none have corrected for STA. The hypothesis was that intact (amputee) and dominant (control) knee angles would differ in gait but not cycling between TT amputee and control participants.

METHODS

Cal Poly's Human Subject Committee approved all protocols. Healthy control (BMI = 23.3 ± 2.6 ; $n=7$, aged 18-29), and TT (BMI = 24.33 ± 1.3 , aged 18-37) participants were screened for abnormalities (e.g. history of knee injury) that could bias biomechanical results. An enhanced Helen Hayes marker set with 32 markers and 2 clusters of 7 randomly spaced markers on the thigh and shank were placed on the participants. Kinematic data were captured using a 12-camera motion analysis system with Cortex software (Motion Analysis Corp., CA, USA). Kinetic data were collected using 3 ground force plates (AMTI, MA, USA) for gait experiments and a stationary bike (Lifecycle GX, Life Fitness, IL, USA) retrofitted with custom pedals with 6-axis load cells (AMTI). A static trial capture was used to identify marker reference

configurations. Gait and cycling experiments were conducted at self-selected walking speeds and at 70 RPM with a moderate machine resistance level, respectively. Kinematic and kinetic data were processed in Cortex and filtered (4th order Butterworth, cutoff frequency 6 Hz), and then exported to MATLAB (MathWorks, USA) for formatting.

A floating axis coordinate system [8] defined knee angles: flexion/extension (FE), internal/external (IE) rotation, adduction/abduction (AA). The 7-marker clusters were used to obtain STA corrected body segment orientations. For every time point, each cluster was partitioned into sub-groups of 3 markers. Then, using a director vector approach, each subgroup was modeled as a Triangular Cosserat Point Element (TCPE). In accordance with [9], metrics quantifying TCPE rotations, deformations, and translations were used to establish a filtering algorithm to select the TCPE's undergoing the least amount of STA. From the selected TCPEs, STA-corrected thigh and shank rotation tensors were determined, and corrected thigh and shank orientations were determined by rotating each of the respective bone segments in the static configuration. Results were normalized to 101 time points corresponding to 0-100% of a gait/cycling cycle. Custom MATLAB codes were used to perform TCPE and statistical analyses. Maximum/minimum knee angles were compared with 2-way ANOVA with post-hoc Tukey tests ($p < 0.05$ significant).

RESULTS AND DISCUSSION

There existed no differences for FE, IE, and AA angles due to subject status (control-TT) or exercise type (gait-cycling) (Fig. 1). Further analysis of the data revealed that there existed a difference ($p=0.006$) between the timing of the minimum adduction angle which occurred at ~50% and 66%

of the gait cycle for TT and control participants, respectively (Fig 2).

CONCLUSIONS

The results did not support the hypothesis that knee angles would differ in gait but not cycling between TT amputee and control participants. Differences in kinematics between TT amputee-intact and control-dominant knees may be found in metrics other than maximum angles (e.g. the timing and asymmetry of flexion angles in both mid-stance and swing phases) as reported in [10]. Thus, future work should include investigating the timing and range of 3-D knee angles that are corrected for STA errors using the TCPE method.

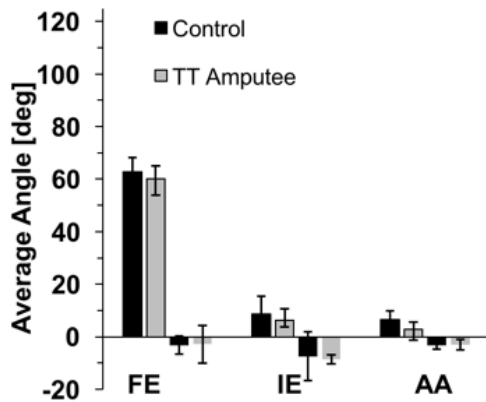


Figure 1. TCPE corrected maximum/minimum FE, IE, and AA angles in gait and cycling averaged over TT (n=7), and Control (n=7) participants. Mean $\pm$ 1 S.D. shown. * indicates significant difference.

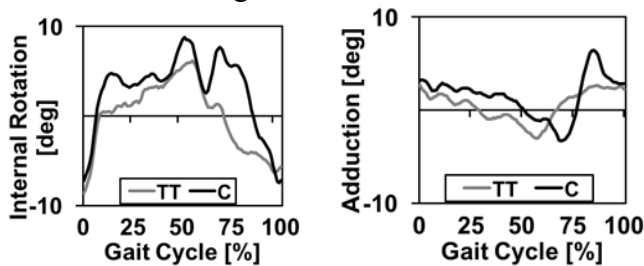


Figure 2. TCPE corrected IE (left) and AA (right) angles in gait averaged over TT (n=7), and Control (n=7) participants.

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SUBJECT-SPECIFIC FINITE ELEMENT MODELS OF THE HUMAN KNEE FOR
TRANSTIBIAL AMPUTEES TO ANALYZE TIBIAL CARTILAGE PRESSURE FOR
GAIT, CYCLING, AND ELLIPTICAL TRAINING

A Thesis
presented to
the Faculty of California Polytechnic State University,
San Luis Obispo

In Partial Fulfillment
of the Requirements for the Degree
Master of Science in Biomedical Engineering

by
Jonathon Stearns
March 2020

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COMMITTEE MEMBERSHIP

TITLE: Subject-specific Finite Element Models
of the Human Knee for Transtibial
Amputees to Analyze Tibial Cartilage
Pressure for Gait, Cycling, and Elliptical
Training

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ABSTRACT

Subject-specific Finite Element Models of the Human Knee for Transtibial Amputees to Analyze Tibial Cartilage Pressure for Gait, Cycling, and Elliptical Training

Jonathon Stearns

It is estimated that approximately 10-12% of the adult population suffers from osteoarthritis (OA), with long reaching burdens personally and socioeconomically. OA also causes mild discomfort to severe pain in those suffering from the disease. The incidence rate of OA for individuals with transtibial amputations is much than average in the tibiofemoral joint (TF). It is well understood that abnormal articular cartilage stress, whether that be magnitude or location, increases the risk of developing OA. Finite element (FE) simulations can predict stress in the TF joint, many studies throughout the years have validated the technology used for this purpose. This thesis is the first to successfully validate a procedure for creating subject-specific FE models for transtibial amputees to simulate the TF joint in gait, cycling and elliptical exercises. Maximum tibial cartilage pressure was extracted post-simulation and compared to historical data. The body weight normalized contact pressure on the tibial articular cartilage for the two amputee participants was larger in magnitude than the control participant in all but the medial compartment in cycling. Additionally, cycling exercise produced the smallest values of contact pressure with elliptical and gait producing similar max values but different areas of effect. The results from this thesis align with the body of work preceding it and further the goal of a FE model that predicts in-vivo articular cartilage stress in the TF joint. Future studies can further refine this methodology and create

additional subject-specific models to allow for a statistical analysis of the observed differences to find if the results are significantly different. Refining the methodology could include investigating the full effect of the damping factor on contact pressure and exploring alternative methods of mesh generation.

Keywords: FEA, ABAQUS, TIBIOFEMORAL JOINT, SUBJECT-SPECIFIC, MESHING, AMPUTEE, OPEN-SOURCE, CARTILAGE

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1. INTRODUCTION

1.1. Motivation

Osteoarthritis (OA) is a degenerative joint disease that occurs most frequently in the hands, hips, and knees. One working definition of OA is a collection of conditions that lead to symptoms associated with reduced structural integrity and mechanical performance of articular cartilage (AC) in addition to related changes in the underlying bone.[1]. It is estimated that approximately 10-12% of the adult population suffers from OA, with long reaching burdens personally and socioeconomically. OA causes mild discomfort to severe pain in those suffering from the disease, a personal economic burden can be as high as \$7,500 per year in medical costs and lost wages. A 2015 study found that adults with OA incurred \$318.4 billion in health care costs and suffered \$10.1 billion from lost wages. A comparison of the health costs of adults with and without OA estimates the direct costs of OA to be \$45 billion in health care costs and \$1.7 billion in lost wages [2]. Aside from the economic standpoint, the quality of life of these individuals is impacted in a way that prevents many from leading the lifestyle they wish to. There is currently no cure for OA, according to the Centers for Disease Control and Prevention (CDC) most physicians will recommend several therapies to treat OA symptoms, most commonly these treatments are physical activity therapies to activate biological signals for the health of AC [3]. With an approximate 30.8 million Americans suffering from OA in 2015, a population largely composed of those over the age of 45 and military veterans, preventative measures that can help to mitigate the number of people suffering from this disease are of extreme interest [4], [5].

This thesis finalizes the work done by the Cal Poly Human Motion Biomechanics (HMB) lab for a project funded by US Army Medical Research and Materiel Command that addresses the problem of decreased quality of life due to the high prevalence of osteoarthritis for military transtibial (TT) amputees. Specifically, the research conducted with funding from this grant addresses the lack of studies aimed at measuring and predicting intact limb knee and hip joint and articular cartilage tissue loads in transtibial amputees using prostheses during common fitness sustainment exercises [6]. The exercises analyzed in this thesis are gait (walking), stationary cycling and elliptical training. Since abnormal loading of the articular cartilage contributes to the development of early-onset OA, it is valuable to determine if these common fitness sustainment exercises have significantly different impacts on articular cartilage loading. This is the primary goal of this thesis.

A secondary goal of this thesis is to develop a robust, repeatable and semi-automated schema for developing subject-specific finite element (FE) models of the human tibiofemoral joint. The modelling schema should be replicable for individual researchers and for various subject-specific geometries obtained from MRI scans. If such a schema can be created, it would greatly enhance the ability of researchers to explore the effects of various conditions that are currently correlated to early-onset OA in the TF joint on the cartilage without invasive methods. The models are required to be robust enough to accurately and consistently output results for various exercises.

It is known from various cohort studies [5], [7] that TT amputees have a significantly higher incidence of knee (TF) OA than non-amputees. This OA is a degradation of the articular cartilage attached to the tibial plateau and predominantly occurs in the intact leg

of the TT amputee. The principal belief held for the cause of this accelerated degradation is when there is significant trauma to a lower limb, the human body compensates by loading the native, intact limb in a way not intended and at magnitudes outside of the range of normal activity [8]. This abnormal loading of the TF joint induces a larger contact stress between the femoral articular cartilage and the tibial articular cartilage, the larger stress is tied to the larger strains in the articular cartilage which leads to accelerated fatigue of the tissue [9]. This implies the most direct measurement for OA risk in the TF joint is the contact pressure on the articular cartilage.

The loading of a human knee is a complex problem. The materials involved are highly non-linear, the loading varies in intensity and frequency over the course of a day, and the geometry of the knee is different from subject-to-subject. These three are the main factors in the drive towards creating a methodology for subject-specific numerical analysis of the response to particular loading cases. The Finite Element (FE) method is well suited to address these factors: material properties are defined by the user for each structure, loading cases are set by the user and are easily adjusted, and three-dimensional models can be used to accurately simulate the true shape of a subject's biological structures.

1.2. Prior Work

The FE method has been used extensively in physiological loading cases. FE models can give results comparable to studies with invasive in-vivo methods without disruption of the body's natural physiology. FE theory has been used for over 40 years [10] specifically for loading of the human TF joint itself, with many studies being conducted in recent years.

Studies concerning FE models for the human TF joint vary in scope. These studies encompass everything from material model studies [11]–[14], to generalized methods for the development of models [15], [16] to whole knee FE [17]–[20]. There is a glut of resources available for perusal with varying methods, material definitions and matter of study. What all these studies can agree on is the sensitivity that whole-knee models have to soft-tissue definitions and why it is a principal concern when discussing clinical results [21].

As mentioned previously, this paper is a continuation of a series of papers beginning with Wangerin in [22] which constitute a body of work conducted by the Cal Poly HMB lab in development of FE models of the human tibiofemoral joint. Each of these papers has contributed to the success of this schema and a short summary of each is necessary to understand the choices made in the development of this schema. The five papers can be separated into three groups based off the methodology used for the simulations.

Wangerin [22] began this series of investigations by creating a FE model of the human tibiofemoral joint that utilized joint contact forces from motion analysis data. The study utilized geometry from the Open Knee public domain repository and motion analysis data from a single participant who was physiologically similar to the donor of the tibiofemoral joint. This study aimed to generate a complete validated tibiofemoral joint FE model for predicting stress and strain in the articular cartilage during gait. Wangerin states a secondary goal of providing a basis upon which future work can build, a commonality with all papers in this series.

Sylvia [23] and Czapla [24] introduced the process of subject-specific segmentation and used loading and boundary conditions from motion capture experiments conducted

by the Physical Activity Energetics/Mechanics Laboratory at Colorado State University, Fort Collins. Sylvia focused on creating a simulation with a goal of identifying risk factors for early onset osteoarthritis of articular cartilage in obese individuals and individuals with malalignment in the tibiofemoral joint. Czapla focused on identifying risk factors for early onset osteoarthritis of articular cartilage for individuals with ACL injuries and reconstructed ACL bundles. Both theses emphasized the potential clinical application of a highly robust FE tibiofemoral joint model that predicts if a specific activity could lead to early onset OA in at-risk groups.

Lane [25] and Rumery [26] have the most similar methodology to the one discussed in this thesis. The current project can be seen as a direct continuation of the efforts of Lane. Lane saw the benefit of automating the computational meshing process and FE model generation. His success in creating a single robust model from MRIs (Magnetic Resonance Images) for the subject-specific loads and boundary conditions available from the HMB lab at the time was novel. This model was used to compare cartilage contact pressure between transtibial amputee and control loading cases in gait and stationary cycling. However, this was only for one tibiofemoral geometry and for resultant loads across the knee instead of the actual contact force acting at the TF joint. Rumery attempted to address the limitation of a linear elastic isotropic material model for the articular cartilage and validate a single material model that could be used in more complex loading conditions of the tibiofemoral such as in gait, cycling and elliptical training. He tested three different elastic material model: linear isotropic, neo-Hookean and poroelastic. Rumery utilized three different tibiofemoral geometries in this work and compared results against experimental data from a cadaver study.

This thesis has the benefit of these past works to reference while furthering the ultimate goal of developing a comprehensive subject-specific FE simulation that utilizes subject-specific load cases to determine risk factors for early onset osteoarthritis. There are some acute differences to the first three theses, primarily the at-risk group being studied are individuals with transtibial amputations of one leg and entirely subject-specific data. This includes MRI scans of each subject who underwent motion analysis experiments for gait, cycling and elliptical training at the Cal Poly HMB lab. The MRI scans provided the subject-specific tibiofemoral joint geometry while the motion analysis experiments provided joint contact data. Due to the ambitious nature of this goal, a new semi-automated model creation schema was introduced by the last two papers to allow the creation of multiple knee geometries within the scope of a thesis project. These last two papers, especially Lane [25] are very similar to the work in this paper with key differences being the robustness of the methodology and the magnitude of the loading conditions used. However, what distinguishes this thesis from all previous work in the field is the combination of subject-specific geometries with subject-specific loading cases gathered from motion analysis experiments. The loads used are the most physiologically accurate to date through the use of inverse dynamics software that calculates the joint contact force separated from the muscle forces acting at the tibiofemoral joint.

This thesis is novel in part thanks to advances in knowledge and data from the motion analysis portion of the Cal Poly HMB lab that has been detailed through several award-winning conference abstracts such as in [27]–[31]. The loading conditions used in this paper are from the results presented in [32] with motion capture protocols from [33][34]. The other novelty of this paper is the use of three subject-specific knee geometries. While

Rumery also utilized three subject-specific geometries, the models in this current thesis are able to undergo large deformations while simulating the maximum loading cases for gait, cycling and elliptical training exercises and produce consistent results with these extreme loading cases. The extreme loading cases were not able to converge using past methodologies. For these FE simulations to be viable for clinical recommendations they must be able to provide results for the most extreme conditions which is a large accomplishment made in this thesis.

1.3. Objectives

This study investigates the pressure on the articular cartilage of the tibial plateau. There is to be a comparison between three different exercise types (gait, cycling and elliptical training) on the same subject as well as any differences between the control and TT amputee groups for each exercise. This thesis details results from one control participant and two TT amputee participants. These exercise types were chosen for their accessibility as well as the range of loading modalities. Gait represents weight-bearing impact exercise and cycling represents non-weight-bearing non-impact exercise with elliptical training in the middle of the two extremes. Differences between the control and TT amputee subjects can be used to highlight possible increased risk to the TT amputees in the development of OA in the articular cartilage. The parameter measured here to assess increased OA risk is the contact pressure experienced by the articular cartilage while under load in the given exercises. The contact pressure is directly related to the stress in the AC bodies. It is known that too large of a magnitude of stress in biological tissue can cause cellular degeneration which is a large factor in the development of OA.

Cartilage contact pressure is determined through the aforementioned FE models. Three subject-specific FE models are created and subjected to subject-specific loading cases from gait, cycling and elliptical training experiments. The FE models consist of subject-specific geometry obtained from MRI scans conducted after the exercise experiments. Material properties applied to the tissues in the models are assigned based on prior work in the field. These geometries in conjunction with applied loading cases recreate the *in vivo* TF environment during these exercises and produce stress in the tibial plateau articular cartilage. The models first undergo a validation study based off prior work in the field before the desired loading cases are applied.

This study also aims to further develop a methodology for both the Cal Poly HMB lab and the field at large in development of subject-specific FE models for the collection of *in vivo* data of soft tissues in a non-invasive manner. Subject-specific FE models are time intensive to develop and validate. It is in the best interest of the community of researchers in this field to develop an automated methodology for developing the models with documentation validating this process. A secondary motivation for this paper is to present a step towards that goal by detailing a robust semi-automated process for developing a large number of subject-specific geometries that can be combined with subject-specific loading cases to create a subject-specific FE model. This thesis also aims for the process to be accessible to researchers in this field who understand the basics of FE analysis while not requiring large time investiture to understand the minutiae of the entire process.

2. METHODS

This section details the overall process of creating the FE models used in this study. The minutiae of the process are contained in the attached appendices which were created

in accordance with the protocol procedure of the Cal Poly HMB lab. The aim of this section is to give an understanding of the overall process and give justification to the modeling choices made. The Cal Poly HMB lab protocols found in Appendices A-D are meant to detail the exact processes within each software package that was used to precisely replicate this methodology for future work.

The novelty of this study lies in the use of subject-specific loading and boundary conditions being used on the corresponding subject-specific geometry to create unique subject-specific models. Previous studies with the HMB lab have used a single tibiofemoral geometry of one individual and applied to that model loading and boundary conditions from multiple participants [25] or multiple subject-specific geometries with boundary conditions from previous experiments in the field [26], [35].

2.1. Participant Information

A total of three participants underwent the experimental procedures necessary for this study. One (1) was a healthy control participant and two (2) were amputee participants that met the criteria of the research project [6]. Table 2-1 gives the relevant physiological information of the participants.

Table 2-1. Physiological Data Of Participants

Subject ID	Control / Amputee	Dominant Leg	Age	Body mass [kg]	Height [cm]	BMI
2018Jan19-01	Control	R	22	79.1	182	24.0
2016Nov10-01	Amputee	L	32	74.9	182	22.6
2016Nov14-02	Amputee	R	31	83.9	169	29.4

2.2. Motion Capture Experiments and Data Processing

Participants for this study were recruited to participate in a motion capture experiment pursuant to the methodology in [32]–[34]. These experiments utilized a standardized set of marker-based motion capture equipment as well as a either ground force plates or 6-axis load cells. The marker system uses a series of near-infrared digital cameras to capture the location of photo-reflective markers for the duration of the motion-capture experiments. This set uses 32 photo-reflective markers and is a modified Helen Hayes set shown in Figure 2-1 and an experimental example in Figure 2-2 below.

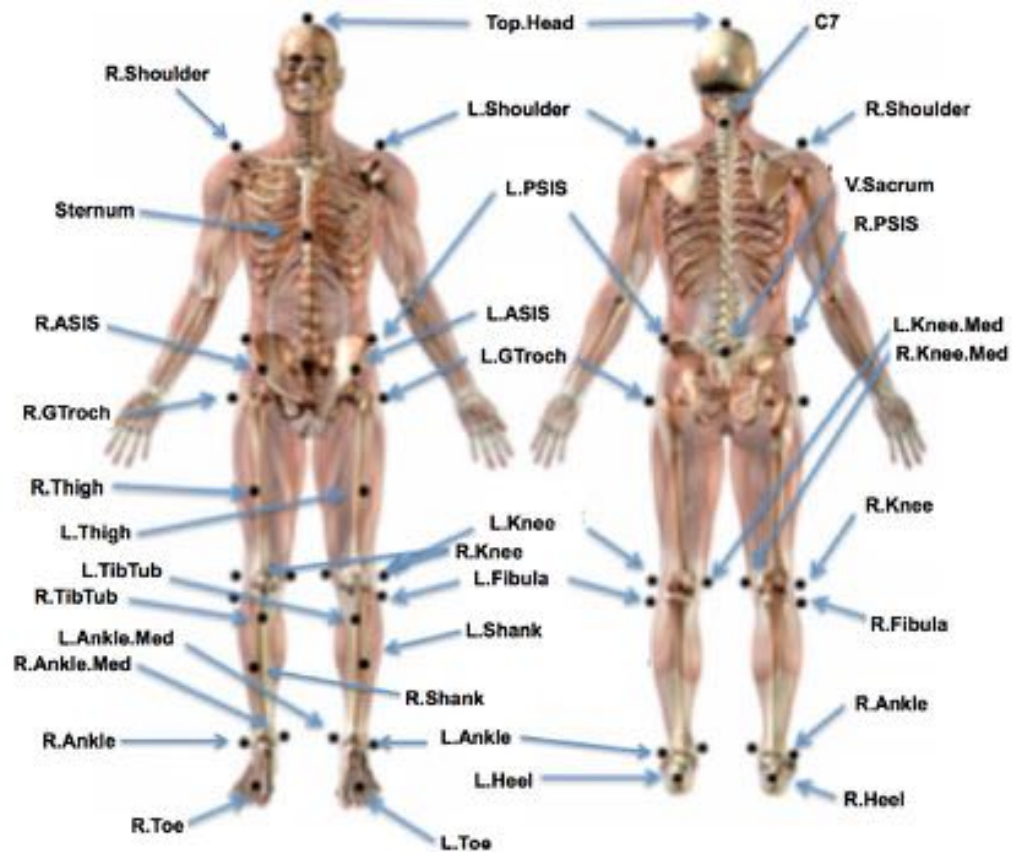


Figure 2-1. Modified Helen Hayes Marker Set Used In Motion Capture Experiments. Reprinted From [36]

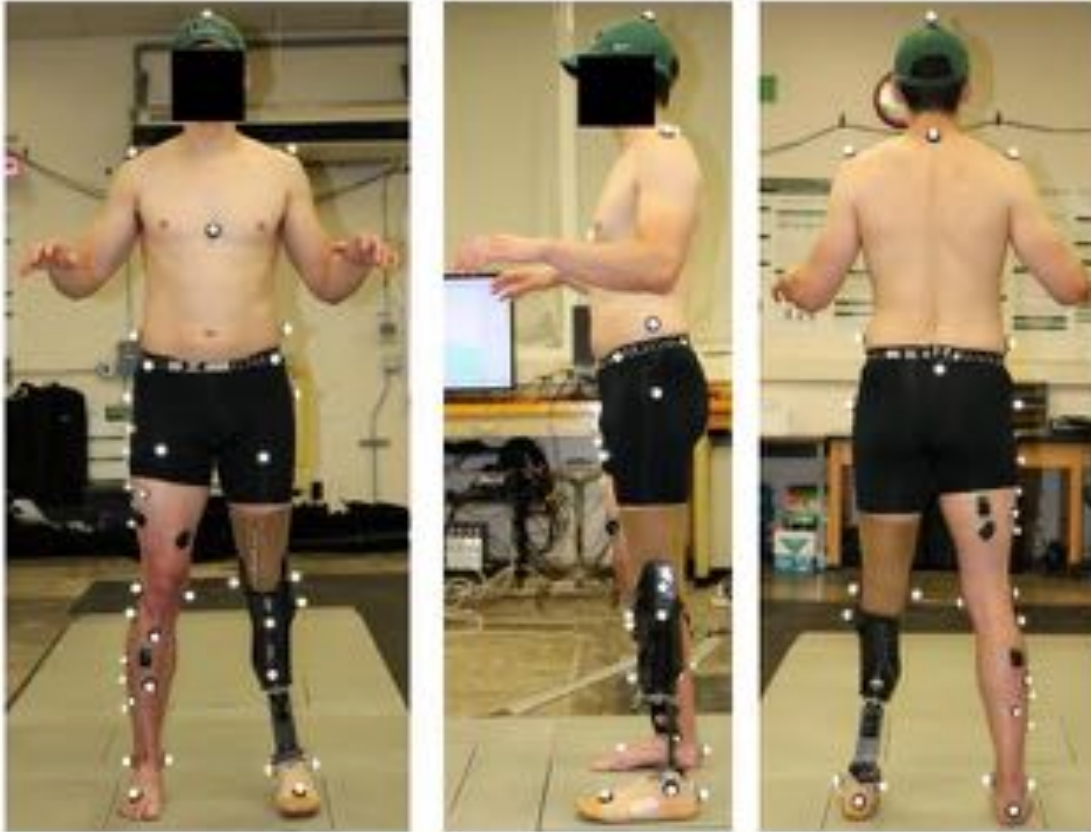


Figure 2-2. Amputee Participant With Modified Helen Hayes Marker Set. Reprinted From [33].

The participants participated in three exercises: gait (walking), stationary cycling and elliptical training. The data gathered from the motion capture system was initially processed in Cortex (Motion Analysis, Rohnert Park, CA) [37] to obtain marker position and force data. Additional processing in the inverse dynamics software OpenSim (Simbios, Stanford, CA) [38] following the protocol presented by Fernandez et. al in [32] allowed for the determination of joint contact forces, part of the loading and boundary conditions in the FE model. The loading data gathered from OpenSim includes muscle forces and joint contact forces. In his study preceding this thesis, Lane [25] utilized forces from Cortex from Orekhov in [33], [34] which are known as joint reaction forces. Joint contact and joint resultant forces at the knee are related, however joint contact forces are typically around 2 to 3 times greater in magnitude as joint resultant forces. This is

because joint resultant forces include all muscle forces acting at the tibiofemoral joint and are not an accurate indicator for the force exerted on the articular cartilage. Joint contact forces separate the force carried by the articular cartilage from the muscle forces and are more indicative of the true loading conditions on the articular cartilage during exercise.

While the process of gathering the loading and boundary condition data will only be briefly discussed in this paper, it is important to note the semi-automated nature of the motion analysis procedure. Currently in the Cal Poly HMB, several templates exist to automatically identify marker data through all frames of motion capture. Marker data also includes a static reference frame to scale the data as it is transferred from Cortex to OpenSim . This scaling process has yet to be automated and must be done by a trained researcher. A motion capture experiment takes approximately 1.5 hours, marker identification in Cortex takes 4.5 hours, scaling in OpenSim takes 10 hours and the inverse dynamics analysis in OpenSim takes 2 hours. The average time spent on a single participant from motion capture to output of knee boundary conditions (loads) is approximately 18 hours. Marker identification is semi-automated, scaling is manual and the inverse dynamics analysis is automated.

2.3. Subject-Specific Geometry Development

Following the motion capture experiments, the geometry for each participant was developed from MRI scans to a final set of 3-dimensional models.

2.3.1. MRI Scans

The MRI scans were performed at French Hospital (San Luis Obispo, CA) on a GE Signa HDxt 1.5T scanner (GE, Chicago, IL) with the same settings for all four scans: proton density fast spin-echo, fat saturated sequence (4800 second relaxation time, 32.1

second echo, 2 averages, 90-degree flip angle) with a 1 mm slice thickness in the sagittal plane within a 512x512 matrix. The scans were all centered on the TF joint and included sufficient lengths of the femur and tibia as determined by the technician to include all relevant ligament and tendon attachment points. These settings were used on the recommendation of a local radiologist with the purpose of making the various soft tissues (cartilage, meniscus, ligaments) distinct enough to segment. The MRI data were anonymized using a script from [25] and in accordance with the protocols in [6] to remove the participants' personal data.

2.3.2. MRI Image Segmentation

For the subject-specific finite element studies, eight bodies in each TF joint needed to be created with their anatomical definitions shown in Figure 2-3: femur, tibia, fibula, femoral cartilage, lateral tibial cartilage, medial tibial cartilage, medial meniscus and lateral meniscus.

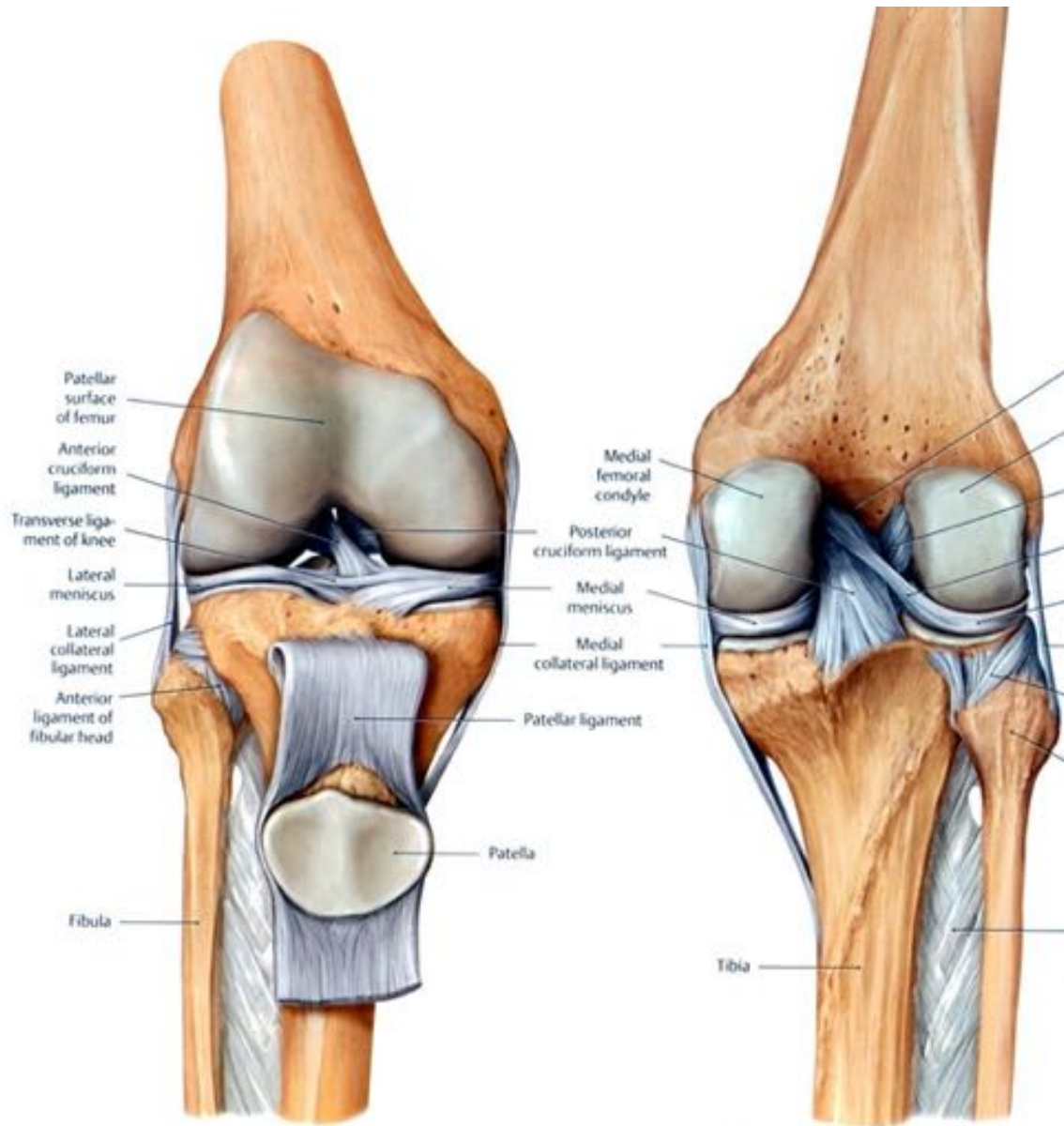


Figure 2-3. Anatomy Of The Human Knee [39]

These bodies were deemed the critical components to acquire physiologically relevant cartilage pressure. Ligaments were not segmented as the MRI scans did not produce sufficient detail of the ligaments to allow for accurate construction. An alternative method for ligament creation is discussed in 2.3.5. The bodies were created through a process known as “segmentation” where the operator utilizes image-processing software to define the boundaries of the bodies in each of the images contained in the MRI

sequence and then combine each of the individual 2D boundaries into a 3D surface. This process was performed manually for all four TF joints using ITK-SNAP (University of Pennsylvania, PA, USA) [40]. ITK-SNAP was chosen as an open-source alternative to commercial solutions such as Materialise Mimics (Materialise, Leuven, Belgium). ITK-SNAP has a low entry-barrier for the manual segmentation process and shows potential for an efficient automated segmentation process. An example of one image in one of the MRI sequences is shown in Figure 2-4 and an example of the 3D surfaces of the TF joint is shown in Figure 2-5.



Figure 2-4. Fully Segmented MRI Sequence Image Of 2016Nov05-01RC. Sagittal Plane View. Femur (Red), Femoral Cartilage (Pink), Lateral Meniscus (Teal), Lateral Tibial Cartilage (Blue), Tibia (Green) And Fibula (Tan).

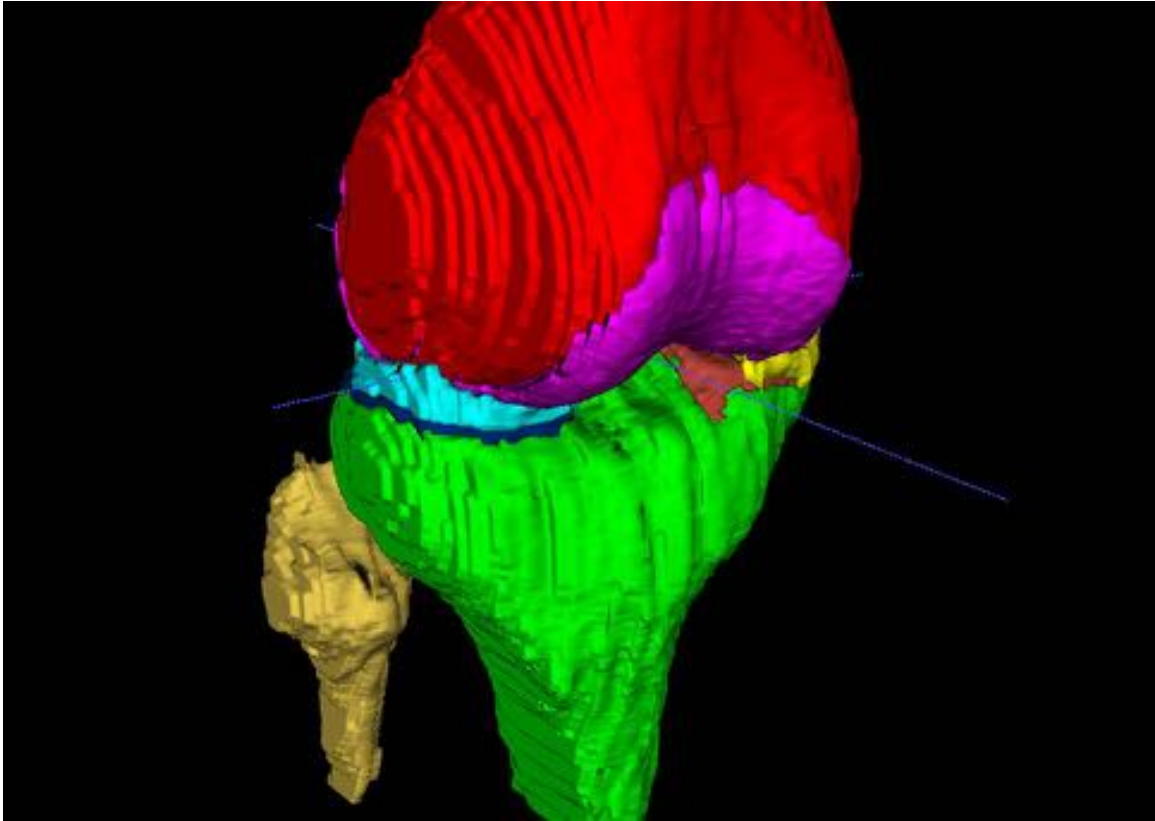


Figure 2-5. 3D Surfaces Of The Bodies In The TF Joint Viewed In ITK-SNAP

The 3D surface that is created by this process is commonly represented by a 3D Piecewise Linear Complex (PLC). ITK-SNAP uses three different file formats to represent a PLC, the stereolithography (STL) file type was the format used to extract the surface definition from ITK-SNAP to continue with processing the bodies for the FEA. The protocol for the manual segmentation of the TF joint is outlined in Manual MRI Segmentation. This process takes approximately 35 hours for each knee geometry for a moderately experienced user. This step is currently the largest limiter in the creation of the FE model with respect to both time and expertise.

2.3.3. Smoothing

The manually segmented bodies have rough surfaces that exhibit staircase artifacts typical of 3D models generated from MRI or CT scans due to the low resolution in the

direction normal to the scan [41]. This is shown clearly in Figure 2-5. The exported 3D surfaces also have extremely large vertex and face counts resulting from the segmentation method. These two factors mean the raw 3D surface-bodies are not suitable for FE meshing, the mesh is much too dense to be computationally viable and the sharp edges of the staircase artifacts cause a distortion of elements.

The bodies were refined using the mesh processing tool MeshLab (ISTI - CNR, Pisa, Italy) [42] to represent the true anatomy more accurately and allow for FE meshing. MeshLab was chosen for this processing step because of its open-source nature and the wide gamut of filters available in the program. The general workflow of the smoothing and face reduction in MeshLab is shown in Figure 2-6 and a comparison of a mesh pre-smoothing and post-smoothing is shown in Figure 2-7.

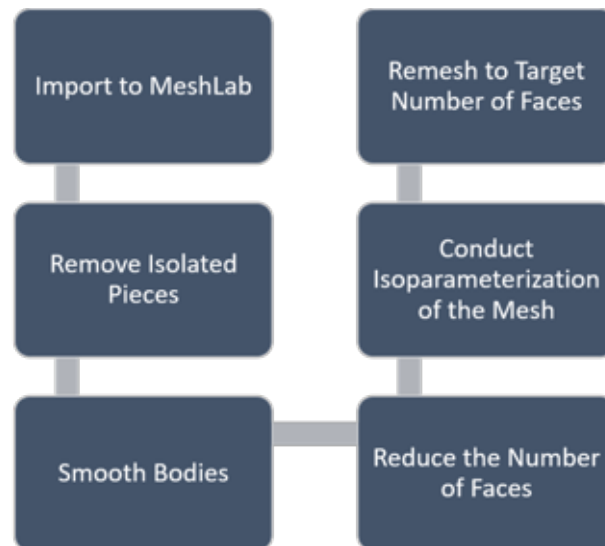


Figure 2-6. Smoothing Workflow Of Meshlab (ISTI - CNR, Pisa, Italy).

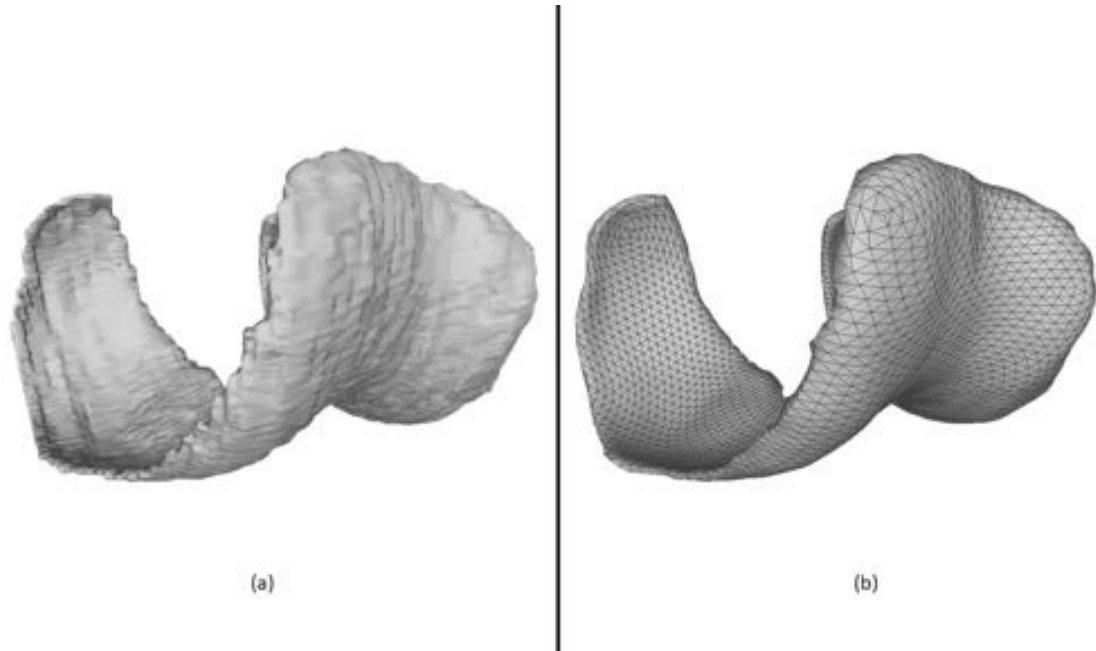


Figure 2-7. Before (A) And After (B) Smoothing And Face Reduction Of The Femoral Cartilage Of
2016Nov05-01R

After each step in the workflow, the mesh volume was calculated. This was used to validate the accuracy of the models as the smoothing and face-reduction workflow caused a reduction in structure volume at various steps. This reduction in volume is unavoidable as the process naturally removes sharp edges through an averaging process. The smoothing process uses the Taubin two-step formulation [43] which involves two successive Gaussian steps to minimize the natural shrinkage of the body. The Gaussian method of smoothing adjusts node locations based off a scaled weighted average of the neighboring nodes, the Taubin two-step process uses two scaling factors λ and μ . λ is used for the first Gaussian step and is a positive number between 0 and 1. μ is used for the second Gaussian step and is a negative number between 0 and -1. The effect of a second step with a negative scaling factor causes the mesh to slightly grow instead of shrinking through the Gaussian smoothing procedure. Through experimentation, the relationship between λ and μ can be set to minimize the amount of shrinkage of the mesh.

The soft bodies (cartilage and menisci) were much more sensitive to the smoothing and face reduction process than the bones due to the thin cross-section and the large curvatures present. All bodies used the same scaling factors of $\lambda = 0.60$ and $\mu = -0.59$, however the soft bodies underwent 300 smoothing steps while the bones underwent 800. The reduction in volume of most of the soft bodies was kept to less than 5% of the original volume while the reduction of volume in the bones was kept to less than 2%, often being less than 1%.

The smoothing procedure removed staircase artifacts but does not address the extreme number of faces. The number of faces was reduced through the Quadric Edge Decimation filter in MeshLab. The target number of faces for each body was set to be 8000, this was determined experimentally to be the most efficient for the methodology utilized in this thesis as it resulted in a negligible change in volume for all bodies without excessive time spent minimizing the face count for each body (32 in total). The details on the settings of the filter are described in B. Geometry Smoothing.

With the bodies smoothed and the faces reduced, the bodies were considered viable for FE meshing. However, the triangular facets that compose the bodies are irregular in size, form and location. These irregularities make a computational mesh ineffective, especially when it comes to refinement of a mesh. To create a more efficient mesh, the bodies were re-meshed using an isoparametric filter [44]. Isoparameterization maps the surface mesh into an abstract domain by iteratively breaking the mesh into several large triangular facets that are optimized to create more regular mesh boundaries. Once the initial abstraction is accomplished, a second filter is applied that subdivides these large facets into smaller triangular face that match the existing topology of the

surface. This process is not entirely predictable, as the optimization of the abstract domain uses an unstable formulation that is highly dependent on starting criteria. The instability of the process can lead to areas in the re-meshed bodies that are highly irregular, create holes in the body, or cause the triangular facets to intersect. These issues make the construction of a computational mesh impossible. It remains unknown the exact causes of these issues, generally it can be rectified by reapplying the initial isoparameterization filter to the smoothed body. The abstract mesh parameters of minimum and maximum number of faces directly affect the stability of the final mesh, this input was altered for each body as needed but was always between a minimum of 1 face and a maximum of 100 faces.

The bodies were then re-meshed at various levels of optimization and exported as an ASCII text STL file to be converted into either a volume or surface computational mesh compatible with Abaqus (Dassault Systemes, Vélizy-Villacoublay, France).

The smoothing and remeshing of the surfaces are semi-automated processes. MeshLab has a built-in function to save and open scripts that allow for filters to be applied in a single click with user-chosen parameters for each filter already defined. Each knee takes approximately 30 minutes to smooth and prepare for computational meshing.

2.3.4. Creation of Computational Mesh

The computational mesh of each TF joint contains two distinct categories of bodies: bones and soft-tissue. Bones can be represented by rigid surfaces for computational efficiency as the stiffness of bones is much larger than that of the soft-tissue [45], [46]. The soft-tissue must be represented as a continuum to gather accurate data on the stress and strain of the bodies during exercise. The nature of open-source software leads to

software being better at one function than another, in this case the rigid surfaces were meshed using GMSH (GPL, Liège, Belgium) [47] while the continuum bodies were meshed using TetGen(WIAS, Berlin, Germany) [48], [49] through the MatLab (MathWorks, Natick, MA) extension GIBBON [50]. TetGen was chosen for its ability to be used with MatLab and for the continuum meshing to be fully automated. GMSH was chosen for its capability of surface meshing. Both programs are fully open-source and output files compatible with Abaqus FEA. The procedure for the creation of the computational mesh can be found in Computational Surface Meshing for the surface bodies and D. Computational Continuum Meshing for the continuum bodies.

The creation of the rigid surfaces representing the bones was straight-forward with GMSH. The smoothed surface output by MeshLab was the same surface being used in the Abaqus FE model. GMSH served as a simple way to transcribe the .stl file from MeshLab into the Abaqus FEA native .inp format.

TetGen allows for control of the meshing algorithm through string switches, the string used for the meshing in this study is “-pYq1.2/15Oo/150V”. Table 2-2 below lists each of these parameters and their effect on the computational tetrahedral mesh generation, also repeated in D. Computational Continuum Meshing. These switches have full definitions and thorough explanations of all sub-options in the TetGen user guide [49].

Table 2-2. Parameters Used For Generation Of Continuum Mesh With Tetgen

Switch	Explanation and Chosen Parameters
--------	-----------------------------------

p	Reads a 3D PLC (Piecewise Linear Complex) and generates a tetrahedral mesh.
Y	Used alongside -p. The boundary edges and faces of the input PLC are preserved in the generated volume mesh.
q	Adds new points to improve the mesh quality based off two numerical constraints separated by a “/”. First constraint: Maximum radius-edge ratio. Default is 2.0. Second constraint: Minimum allowed dihedral angle. Default is 0.
O	Specifies mesh optimization using two integers separated by a “/”. If no integers are specified than the default is used. First constraint: Mesh optimization level from 0-9, 9 being max. Default is 2. Second constraint: Allows certain operations such as edge/face flip, vertex smoothing, vertex insertion/deletion. From 0-7, 7 being all operations allowed. Default is 7.
o/	Alters the objective function of Tetgen. The objective function of Tetgen 1.4 is to reduce the maximum dihedral angle of the tetrahedra. The number after the switch alters the value of the allowable angle. Default is 165 degrees.
V	Outputs detailed information about the generated mesh to quickly check that the various options are behaving as expected.

The choice of these switches must be explained, as they directly affect the computational mesh, the first building block of a finite element study. The accuracy and therefore the relevance of a FE study is based off the computational mesh because it defines the geometry actually being used in the simulation. The mesh should closely approximate the real geometry of the bodies in question and have a sufficient number of degrees of freedom for the underlying mathematics to converge to a consistent solution.

The “p” switch is required as the input for the continuum mesh is the PLC surface that defines the geometry. The PLC surface used is the STL file output by MeshLab. The “Y” switch is used to retain the isoparametric surface created in MeshLab for consistent meshes to be created for both a mesh sensitivity study on a single knee assembly and for the creation of subsequent knee assemblies. The “q” switch and its sub-options were discovered to be necessary to reduce the number of invalid (zero-volume) and deformed elements generated by TetGen, this is also true of the “O” switch. The “o/” switch allows for better mesh creation by directly altering the objective function of TetGen, similar to the 2nd option of the “q” switch but for a maximum dihedral angle instead of a minimum. The “V” switch is not necessary for the generation of the mesh, instead it is helpful for

troubleshooting various issues with generation of a mesh suitable for FE studies. TetGen does not output a mesh that is immediately ready for analysis, a further step to prepare each body for analysis is discussed later in 4.2.3.

Abaqus FEA has a variety of element definitions that can be chosen based off the needs of the simulation. The methodology in this study lends itself towards triangular surface elements and tetrahedral continuum elements. For the surface meshes (femur, tibia, fibula) the R3D3 linear triangular surface element was chosen, as it has excellent compatibility with contact formulations. For the continuum meshes, the C3D10HS quadratic tetrahedral element was chosen as C3D4 linear tetrahedral elements are known to overestimate the stiffness of continuum structures compared to actual structures [51]. The C3D10HS element is a “hybrid element” that has integration points located at the nodes of the element, this improves surface stress results as there is no interpolation error such as in the standard C3D10 element. Refer to the Abaqus FEA User Guide that comes with the installation of Abaqus FEA for a detailed explanation of the behavior of these element types

In most of the FE models discussed in the Prior Work section (section 1.2), hexahedral elements were used for the soft tissue bodies and triangular elements were used for the rigid bodies, with hexahedral elements generally being regarded as more accurate than tetrahedral elements [45], [46], [51]. However recent studies have shown that in contact analysis, Abaqus FEA quadratic tetrahedral elements perform similarly to linear hexahedral elements in evaluation of contact pressure[52]. Tetrahedral elements generally conform better to geometry that exhibits a large amount of curvature which includes all soft tissue bodies in the TF joint. They are also less computationally intensive

compared to quadratic hexahedral elements; however quadratic tetrahedral elements are less numerically robust than quadratic hexahedral elements. This is discussed later in 4.2.3 specifically when considering convergence of the FE model and the quality of the mesh. Care was taken to ensure that the parameters altering the meshing scheme in TetGen created continuum meshes that had minimal errors with few distorted elements.

The process of meshing with TetGen is fully automated with custom MatLab scripts. The meshing process with GMSH is manual, but the simplicity of the process makes the invested user time minimal. It takes approximately five (5) minutes of user time for the computational meshing of a single knee. This is the largest time saving component of the FE model creation. Hexahedral meshing of bodies such as the femoral cartilage is difficult with traditional methods and not accessible with open-source software, commercial solutions do exist.

2.3.5. Abaqus Assembly Development in CAE Module

Once all bodies required for the knee assembly had been converted into Abaqus FEA compatible computational meshes with the relevant procedure, the individual computational meshes were imported to Abaqus FEA and assembled into a physiologically relevant model starting in the CAE visual module. This model includes five soft-tissue bodies (femoral cartilage, medial meniscus, lateral meniscus, medial tibial cartilage, lateral tibial cartilage), three bones (femur, tibia, fibula) and four spring elements that make up the major ligaments of the TF-joint (ACL, PCL, ACL, LCL).

Each of these components requires a material model within Abaqus FEA. As discussed before, the bones are formed from R3D3 elements which does not allow direct

assignment of material properties. The element definition forces the body defined by the surface to be computationally rigid, infinite stiffness in physical terms.

The ligaments were formed from CONN3D2 elements. These are specialty elements in Abaqus FEA and can be used to simulate a wide variety of mechanical connections such as slots, hinges, bushings. In this case, they acted as perfectly axial springs where force was applied along the line of action that was defined by the connected nodes. This force was based off the relative displacement between the two connected nodes. A CONN3D2 element requires several parts to be fully defined: two node locations, a section assignment, and a behavior assignment. Table 2-3 below describes the necessary components to fully define the CONN3D2 element while Figure 2-8 and Figure 2-9 show how these definitions were depicted in the Abaqus FEA input file. The CONN3D2 elements were defined in the assembly level of the Abaqus FEA model to allow for connection between the different bodies (called instances in Abaqus FEA). In Figure 2-9 a non-linear spring constant is defined. The spring constant is described with displacement and force pairs, which can be read: “For every X units of distance between the nodes, X force units are applied on the line of action between the nodes”. The distance between nodes is the shortest possible straight line and is in millimeters while the force is defined in newtons for this model. The behavior of the non-linear spring allows for pre-tension that exists in ligaments and does not allow for the ligaments to resist any compressive force. However, the spring in Figure 2-9 does not represent a spring with pre-tension, as pre-tension in the spring elements caused convergence issues with the FE model in Abaqus FEA.

Table 2-3. Components Required By The CONN3D2 Element

CONN3D2 Components	Description
Connecting Nodes	Two nodes connected together with the relative motion between the two nodes altered by the CONN3D2 element.
Section Assignment	Describes in which degrees of freedom the connected nodes will be limited.
Behavior Assignment	Describes how the limited degrees of freedom will impact the connected nodes when relative between the nodes occurs

```

** LIGAMENT ELEMENTS *****
*Element, type=CONN3D2
1, FEMUR.3761, TIBIA.2608
*Connector Section, elset=ACL, behavior=ACL
Axial,
*Elset, elset=ACL
1,
**
*Element, type=CONN3D2
2, FEMUR.2193, TIBIA.5410
*Connector Section, elset=PCL, behavior=PCL
Axial,
*Elset, elset=PCL
2,

```

Figure 2-8. Definition Of Two CONN3D2 Elements In Abaqus Input File.

```

1 *Connector Behavior, name=ACL
2 *Connector Elasticity, nonlinear, component=1
3 0, -100
4 0, 0
5 1493.086475, 100
6 *Connector Behavior, name=PCL
7 *Connector Elasticity, nonlinear, component=1
8 0, -100
9 0, 0
10 1477.191856, 100

```

Figure 2-9. Definition Of CONN3D2 Behavior

This material definition follows the methods and reasoning from [25], [46] and calculates properties directly from [53]. Contrary to Lane in [25], one spring each was used for the ACL, PCL, MCL and LCL ligaments instead of using the physiologically accurate number of bundles. This was done to simplify model creation and due to the lack of clarity in the MRI scans for bundle connection points. As shown in [53] the differences between the material properties when modeled as an axial spring are negligible, this allowed for all four ligaments to use the same elastic modulus. This spring definition was simplified to exclude the toe region. The spring constant used was found using the measured linear modulus (345 MPa) of the ligaments in [53], the average cross-sectional area (2.01 mm²) of all ligaments measured in [53], and the measured initial length of each ligament based off the nodes selected in the Abaqus FEA model. This results in all four springs having distinct spring constants. Table 2-4 below lists the spring constants calculated for one of the subject-specific models.

Table 2-4. Ligament Spring Constants For 2018Jan19-01

Ligament	Spring Constant [N/mm]
ACL	13.7
PCL	16.8
MCL	14.1
LCL	11.3

This method is in-line with Hooke's Law when applied to a uniform bar under tension. While a ligament is not uniform in shape, this is the best approximation with true ligament geometry unavailable.

With all physiological components of the FE model defined, methods for controlling the relationship between these parts must be defined. This includes 12 surfaces that define the way bodies contact each other and how some bodies are permanently fixed to each other. These 12 surfaces were used in 5 tie constraints and 6 contact pair definitions. A tie constraint in Abaqus FEA approximated in these models the physiological behavior of cartilage connecting to bone through subchondral tissue. It did not approximate the method of attachment of the menisci to the tibia, but it was used in the FE models to better approximate the true behavior of the tissue within the model. This limitation is further discussed later. The contact pairs define surfaces that were not allowed to penetrate each other. Abaqus FEA handles contact by applying a force to nodes when it is detected that there has been penetration of a node through a surface of a given contact pair set. This force is directly normal to the penetrated surface at the point of penetration. Table 2-5 below lists the pairs. The difference between a master and slave surface is detailed in the Abaqus FEA user manual. In general, the master surface is based off the stiffer body.

Table 2-5. List Of Surface Pairs For Tie Constraints And Contact Surfaces

Tie Pairs	
Master	Slave
Femoral Cartilage (Superior Surface)	Femur
Medial Tibia Cartilage (Inferior Surface)	Tibial Plateau
Lateral Tibia Cartilage (Inferior Surface)	Tibial Plateau
Medial Tibia Cartilage (Superior Surface)	Medial Meniscus (Inferior Surface)
Lateral Tibia Cartilage (Superior Surface)	Lateral Meniscus (Inferior Surface)
Contact Pairs	
Femoral Cartilage (Inferior Surface)	Medial Tibia Cartilage (Superior Surface)
Femoral Cartilage (Inferior Surface)	Lateral Tibia Cartilage (Superior Surface)
Femoral Cartilage (Inferior Surface)	Medial Meniscus (Superior Surface)
Femoral Cartilage (Inferior Surface)	Lateral Meniscus (Superior Surface)
Medial Tibia Cartilage (Superior Surface)	Medial Meniscus (Inferior Surface)
Lateral Tibia Cartilage (Superior Surface)	Lateral Meniscus (Inferior Surface)

There is some overlap between the tied (or attached) surfaces and the contact surfaces, the tibial cartilages and the respective menisci were both tied together and have contact defined. Physiologically, the menisci are attached to the tibia, the femur and the other meniscus through a complex series of ligaments known as meniscal roots or “horns”.

These attachments are partially shown in Figure 2-10 and described in detail in [54]. The

choice to adhere portions of the menisci to the tibial cartilage it rests on is described in more detail in 4.2.4. All surface pairs except for the meniscus-cartilage tie pairs were chosen to encompass all potential areas of contact between the body, an example is shown in Figure 2-11. A representative surface definition for the inferior meniscus surface in the meniscus-cartilage tie pair is shown in Figure 2-12. Four sub-surfaces made up of 6 triangular faces each were chosen to constrain relative displacement between the menisci and the rest of the TF joint, but still allowed for the compression of the TF joint to alter the contact experienced by the menisci and cartilages.

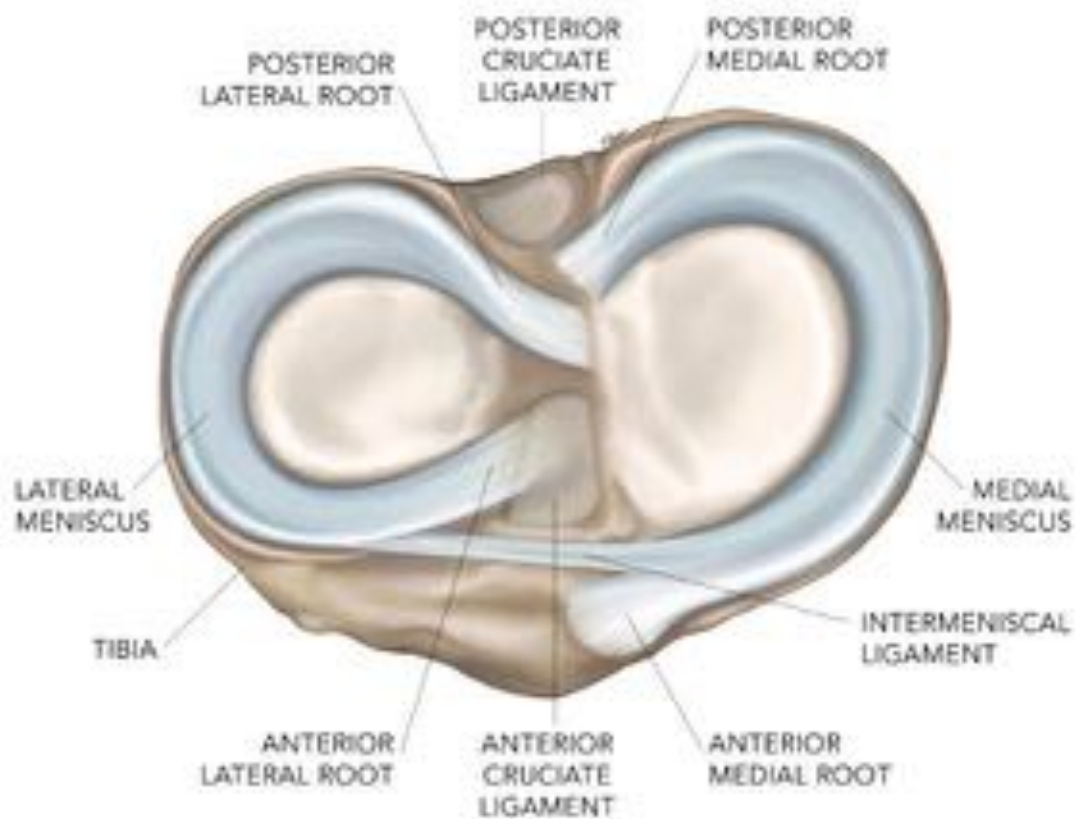


Figure 2-10. Meniscus Physiology [55]

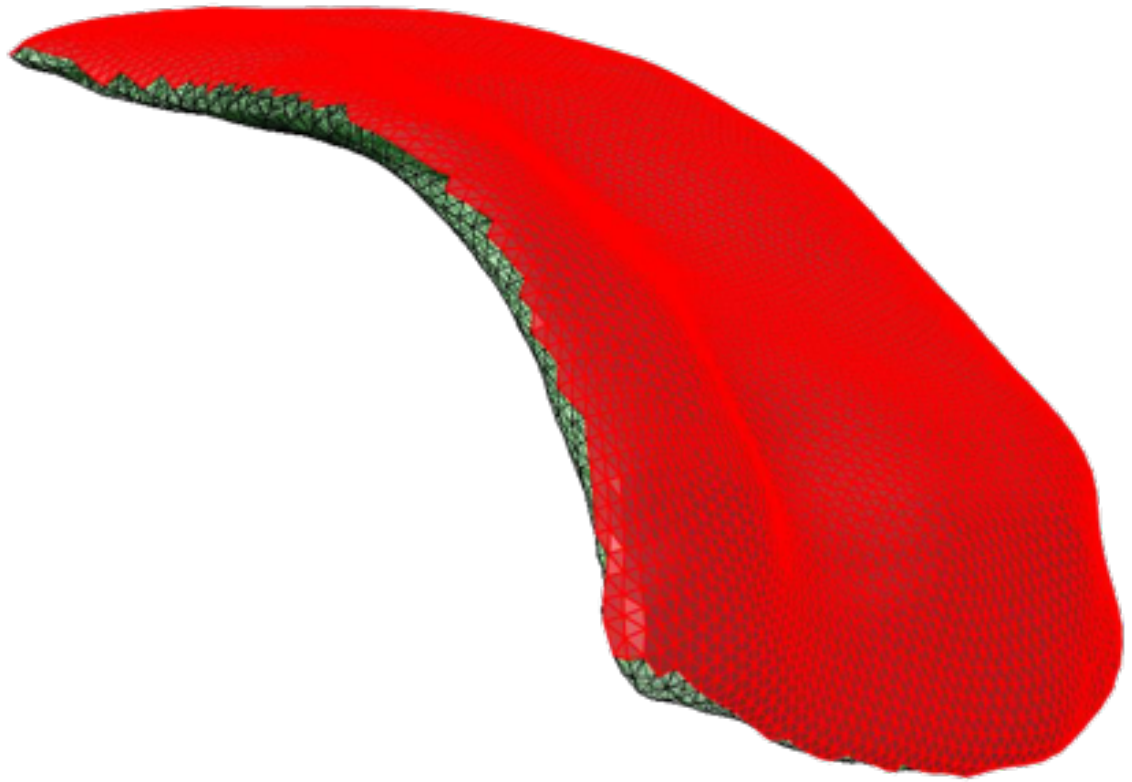


Figure 2-11. Example Of A Typical Surface Definition

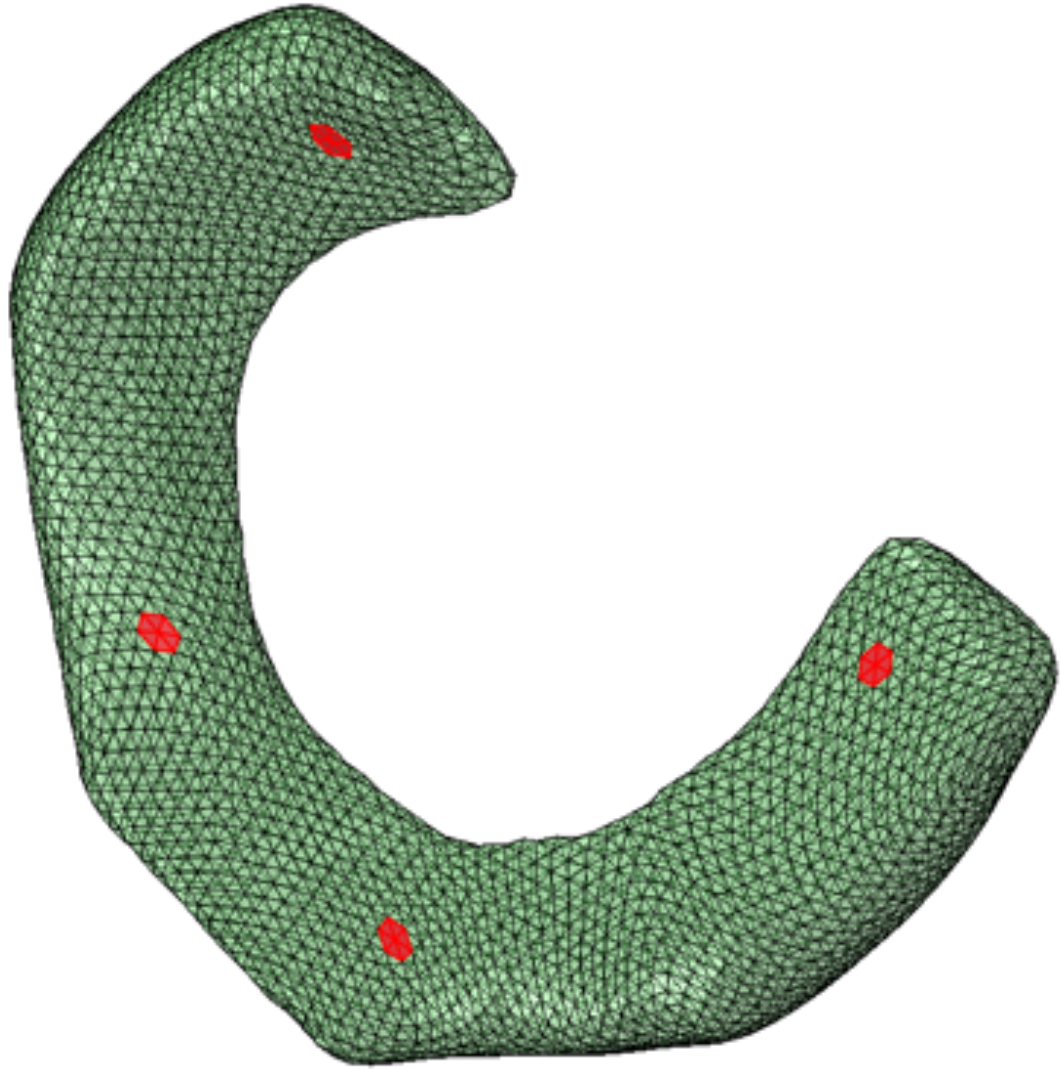


Figure 2-12. Faces Highlighted In Red Define The Surface For The Meniscus-Cartilage Tie Constraint.

The final components for the subject-specific finite element models are the reference points for the rigid bodies. The three rigid bodies in each knee model are the femur, fibula and tibia. The fibula and tibia were paired to the same reference node while the femur had its own reference node. Abaqus FEA requires all rigid bodies to have a reference node where the loading and boundary conditions for the rigid body are applied. The reference node for the tibia-fibula pair was chosen to be the first node defined on the tibia. The location of this node was not critical as the tibia was completely fixed for the

majority of the simulation while the fibula was only used as a reference for the LCL attachment point. However, the simulation was extremely sensitive to the location of the femur reference point. The simulation assumed a quasi-static environment, meaning all loading was assumed from a static tibia reference frame. Further discussed in 2.3.7, there were two steps in the Abaqus FEA simulation: Rotation and Loading.

The location of the femur reference point greatly affected the displacement of the femur during the Rotation step. The femur reference point should be located at the joint center between the femoral condyles for physiologically accurate motion during rotation of the TF joint. In the motion analysis experiment, this was determined by a pair of photo-reflective markers and the joint center was calculated by the motion capture software. In Abaqus FEA, there are several ways to define a reference point. The femur reference point was defined by entering spatial coordinates and manually examining if the point was located on the midpoint of an imaginary line drawn between the most prominent portions of the femoral condyles. Figure 2-13 and Figure 2-14 below represent an accurate placement of the femoral reference point viewed from the sagittal and coronal planes.

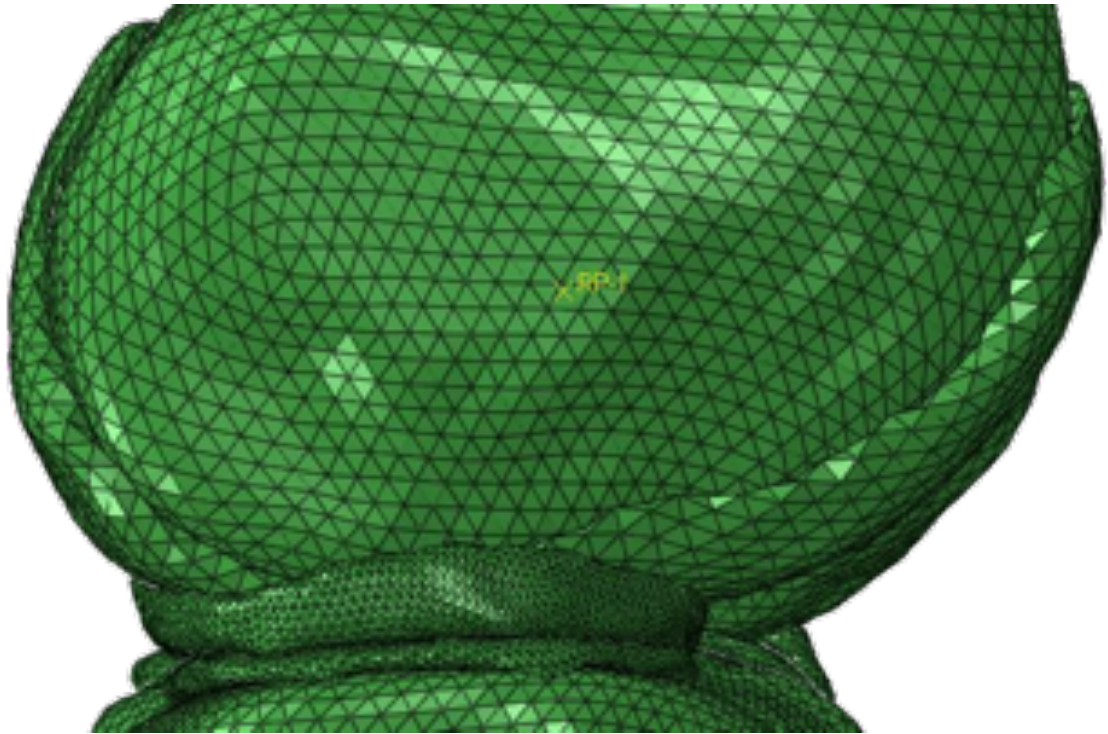


Figure 2-13. Femur Rigid Body Reference Point Of 2018Jan19-01R Viewed From The Sagittal Plane

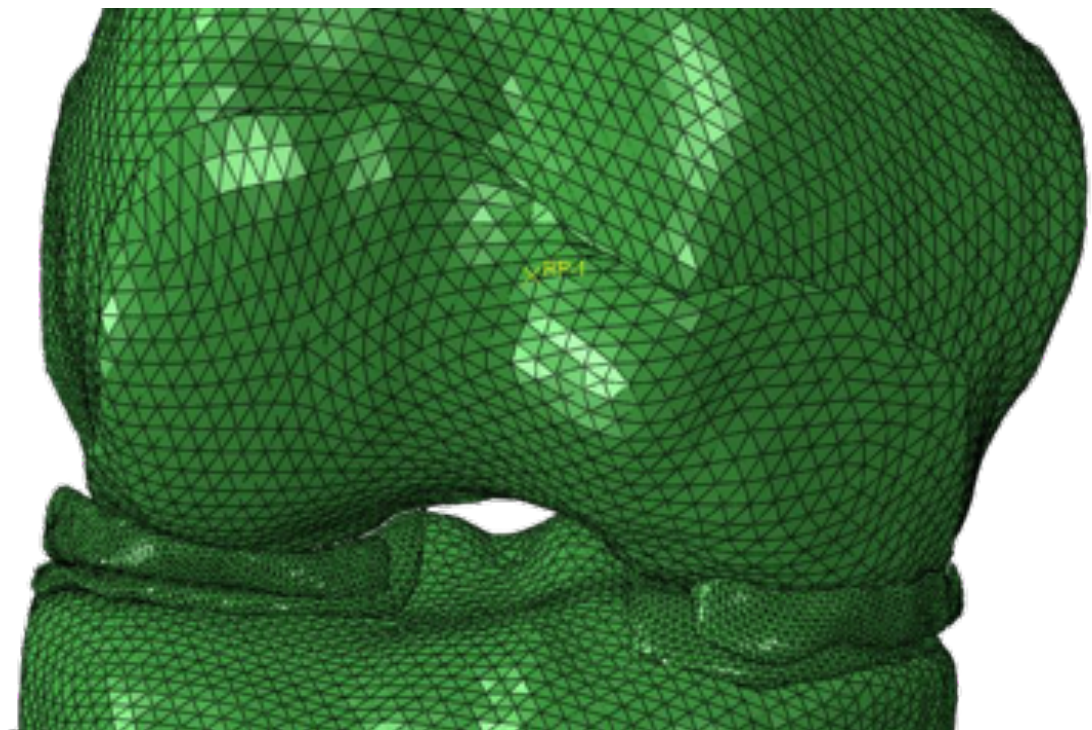


Figure 2-14. Femur Rigid Body Reference Point Of 2018Jan19-01R Viewed From The Coronal Plane

Ideal placement of the reference point would show the minimal displacement at the most prominent portion of both femoral condyles, Figure 2-15 shows a contour plot of total displacement after rotation with a good placement of the reference point.

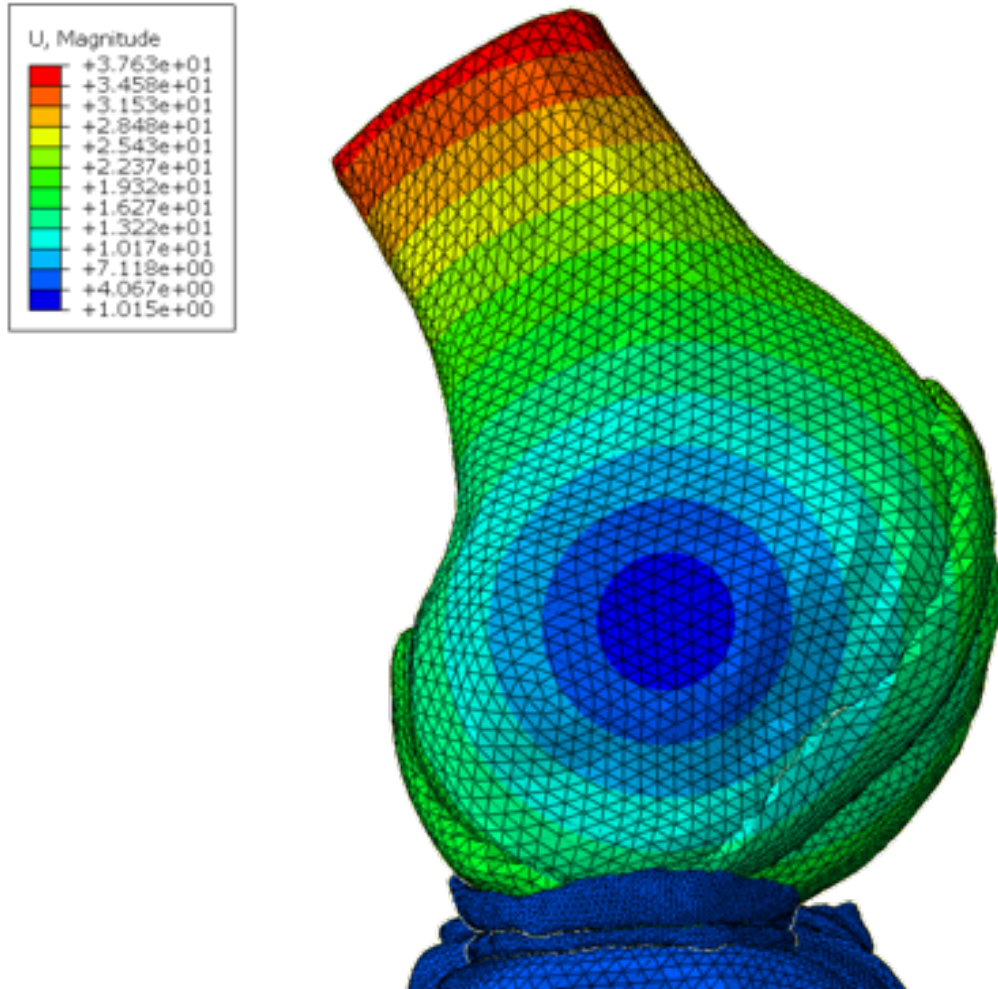


Figure 2-15. Contour Plot Of Total Displacement After Femur Rotation

Once these components of the Abaqus FEA assembly have been defined, the GUI of Abaqus CAE is no longer necessary. The final steps before a simulation can begin involved defining material properties, boundary conditions and other controls for the simulation in the Abaqus FEA input file.

2.3.6. Abaqus Assembly Material Property Definitions

Material properties have already been discussed for the ligaments (section 2.3.5), the non-linear axial springs. Likewise, the bones have been defined as R3D3 elements which have built in material properties of infinitely stiff surfaces. The femoral cartilage, medial and lateral tibial cartilages, and medial and lateral menisci are all continuum elements and require a continuum material assignment. For this study, a linear-elastic isotropic material was utilized. Past work with the HMB lab [26] analyzed the effects of various material definitions on contact pressure in the TF joint. The resulting recommendation was to utilize either a linear or a neo-Hookean material model. Linear material models are the most widely used definitions for cartilage and meniscus properties in FE simulations of the TF joint so a linear material model was used for the cartilage and meniscus definitions following [25], [26], [46]. A summary of all the material models used in the knee model is found in Table 2-6. The linear elastic isotropic material properties for cartilage were calculated from the neo-Hookean material property used by Harris in [56].

Table 2-6. Material Properties Of Knee Model Components

Geometry	Material Model
Cartilage	Linear Elastic Isotropic [56] $E = 41 \text{ MPa}, \nu = 0.495$
Menisci	Linear Elastic Isotropic [46] $E = 59 \text{ MPa}, \nu = 0.48$
Bones	Infinitely Stiff R3D3 Element
Ligaments	Perfectly Axial Nonlinear Springs CONN3D2 Element

2.3.7. Abaqus Simulation Load and Boundary Conditions

Load and boundary conditions for this model included displacement control, velocity control and force control. The TF knee simulation was viewed from a tibial reference frame, all force and displacements were applied to the femur reference point and subsequently the entire femur rigid body. The tibia was fixed in all degrees of freedom for the entirety of the simulation. The first step translates the femur 2mm superior with all other degrees of freedom fixed. The second step the femur was prescribed to rotate about the flexion-extension axis until it matched the angle where maximum compressive force occurred in the exercise cycle, all other degrees of freedom were fixed. The third step the femur was displaced in the inferior direction until contact with the tibial plateau by applying a small force to the femur, the flexion-extension angle was held constant and the axial direction had an applied force while all other degrees of freedom were fixed. The fourth and final step, the femur had the flexion-extension angle held constant while the other rotational degrees of freedom were left free. The femur had the joint-contact forces, based off the motion analysis experiments, applied to the three translational degrees of freedom. Figure 3-16 below outlines the order of the FE simulation steps.

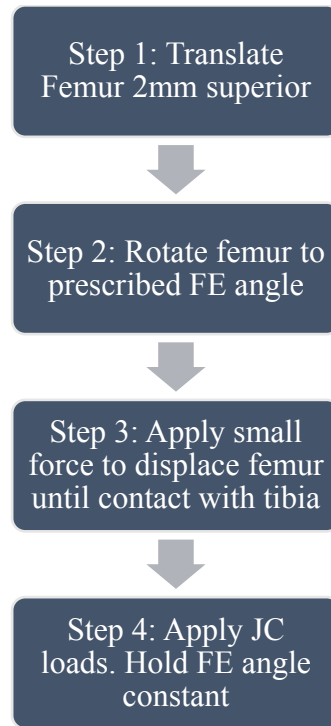


Figure 3-16. Outline Of FE Simulation Steps

Gait, cycling and elliptical training load cases used can be found in F. Subject Specific Load Cases. The load cases were based off the time point in which maximum compressive force occurred. In general, compressive normal force, anterior-posterior shear, and medial-lateral shear joint contact forces were applied. The varus-valgus and intern-external rotation moments were omitted as they were judged unreliable and they also caused convergence issues. In some cases, the anterior-posterior shear and the medial-lateral shear components were also neglected to allow for convergence.

Refer to E. Abaqus Input Files to see how the boundary conditions were implemented in the model.

2.3.8. Abaqus Simulation Controls

When using the subject-specific load and boundary conditions gathered from OpenSim, Abaqus FEA was unable to achieve convergence with the default solver

settings. This was unsurprising given the nonlinear nature of the TF joint models. There were several contributing factors to the nonlinearity: contact sets, tie constraints, nonlinear springs, large deformations, and distorted elements. These nonlinearities were minimized as much as possible while retaining physiologically accurate solutions. According to the Abaqus FEA documentation, it is expected the user will need to alter the convergence criteria to achieve convergence in nonlinear solutions. Several default settings were altered to achieve better convergence including adding global damping, increasing the allowable magnitude for residual forces and displacements, and turning on the line search function. The standard solver for quasi-static simulations in Abaqus FEA utilizes an implicit solver that relies on the Newton method where the solution must be within the radius of convergence for the system. If an increment lies outside of this area, the solution can diverge. By loosening the required limit for residual forces, as recommended by the Abaqus FEA manual for solving highly nonlinear simulations, the incremental solutions can stay within the convergence criteria. Likewise, the damping effect has been shown to stabilize nonlinear systems that include contact and localized damping is recommended for complex contact simulations. The line search function is recommended by the Abaqus FEA documentation to assist in choosing the approximated solution the Newton method requires when solving nonlinear simulations. These alterations are highly dependent on the particular model being simulated and only trial and error can show if they are helpful. In this schema, these parameters, especially damping, helped immensely and allowed for complete convergence in several of the models simulated. See E. Abaqus Input Files for the parameters utilized with the *Controls option.

While damping was required for convergence of the simulations in this thesis, the effects of the viscous forces being applied cannot be ignored. Since applying a global damping factor is commonly used for non-linear simulations, especially with contact, Abaqus FEA has a recommended procedure to check if the viscous forces affect the model in a significant manner. For this thesis, each simulation had the internal energy of the whole model compared to the static dissipation energy for the whole model. Abaqus recommends the ratio of static dissipation energy to internal energy be less than 10% for confidence in physically accurate behavior of a simulation.

The mesh convergence studies as well as the validation study used the automatic stabilization scheme where the global damping coefficient was calculated automatically to ensure the ratio between static dissipation energy and internal energy remained within the appropriate range. The computational simulations required the use of a constant damping factor to converge, this was chosen by trial and error to be .002.

2.3.9. Abaqus Simulation Data Capture

To best approximate a physical measurement of contact pressure on the cartilage, an averaged value for cartilage contact pressure (CPRESS) was measured centered around the location of maximum pressure at the area of interest. This measurement scheme is based off the experimental data from Seitz in [35] where a digital pressure sensor with a spatial resolution of 1.4 mm^2 was inserted between the tibial and femoral cartilages to measure contact pressure. This allowed for a direct comparison in the validation study and the scheme was kept when reporting the contact pressure from the subject-specific load and boundary condition experiments. The node within the area of interest with a maximum value of CPRESS was located and a pattern of 18 nodes adjacent to this point

was created as shown in Figure 2-17. The value of CPRESS at each of the nodes was combined into an average CPRESS value that is the primary reported variable in the study. The area of interest was chosen where the contact pressure was most uniform as FE models can have erroneous stress concentrations that arise from a variety of factors.

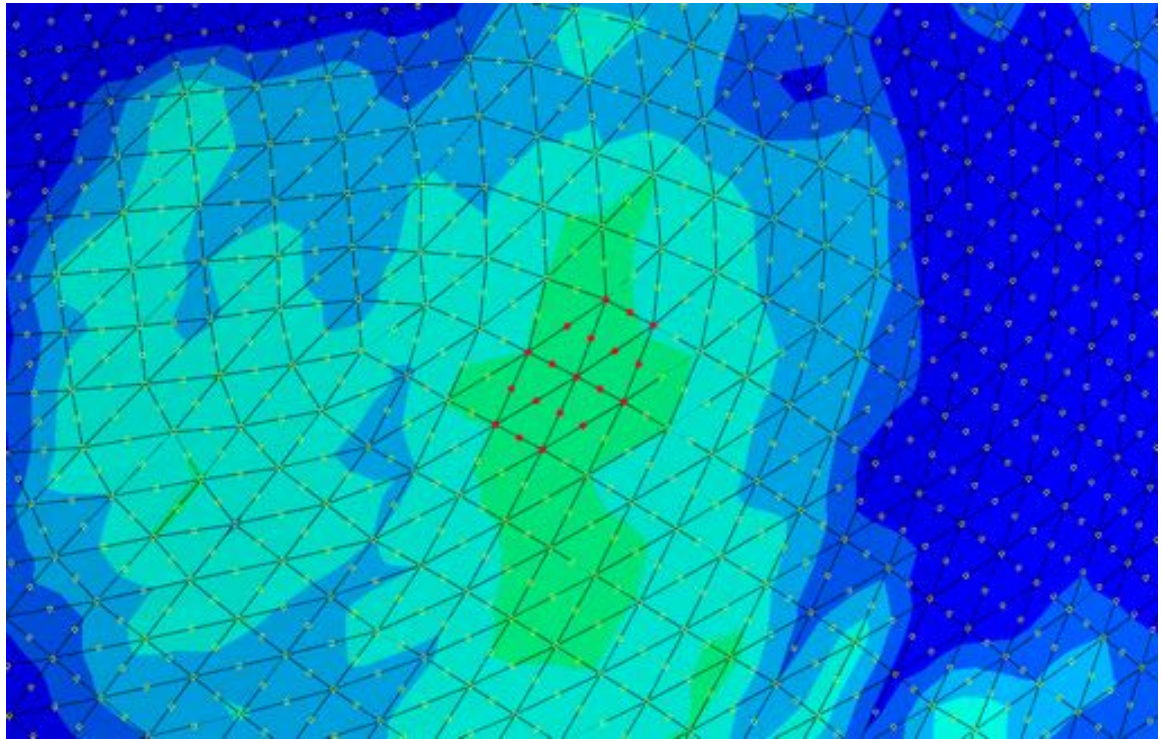


Figure 2-17. Red Nodes Used For Max CPRESS. Centered Node Is The Node With The Max CPRESS On The Body Within The Area Of Interest.

As discussed in section 2.3.8, a secondary variable of interest is the ratio between the static dissipation energy (ALLSD) and the total internal strain energy (ALLIE) of a model. These history variables are calculated for the entire model at each increment. The ratio of ALLSD to ALLIE reported is from the values in the last step of the simulation.

2.3.10. Abaqus Convergence Study

Novel schema for modeling must be validated before the results are useful. The finite element method is a numerical discretization of a 4th order function. The first step to

validate any numerical approximation is to validate the solution has sufficiently small increments, or in this case sufficient degrees of freedom for the error to stabilize. Degrees of freedom in a FE study are generated from a variety of model parameters. The mesh convergence performed in this thesis varied the mesh density of the three articular cartilage bodies. Varying these mesh densities alters the number of elements present in the entire study and thus impacts the degrees of freedom in the study, however this change is not exactly linear as the tie constraints alter the degrees of freedom present in the model and are impacted by the number of nodes interacting between the cartilage bodies and the bones. It was judged sufficient to utilize the number of elements in the individual cartilage bodies for the convergence analysis as the bodies can converge at different mesh densities. For a single mesh body, the number of degrees of freedom is directly related to the number of nodes present which is related to the number of elements and type of element. For this study only C3D10HS (specialized 10-node tetrahedrons) were used for the continuum mesh bodies. The various mesh densities were obtained through the isoparameterization remeshing scheme discussed in 2.3.3. Because this study focusses on the cartilage contact pressure and also because the menisci proved to require a large number of elements to retain their initial volume, a mesh convergence study was performed only through refinement of the three cartilage bodies. Four (4) meshes for each cartilage body of 2018Jan19-01R were created with an increasing number of elements present for each iteration. Table 2-7 below details the number of elements present in each of the successive meshes along with the chosen mesh density for each of the cartilage bodies.

Table 2-7. Number Of Elements In Each Cartilage Body For The Mesh Convergence Analysis. Mesh Chosen For Experimental Simulation Highlighted In Gray

	Mesh 1	Mesh 2	Mesh 3	Mesh 4
Femoral Cartilage	5400	8628	12705	17988
Lateral Tibial Cartilage	6786	10488	15293	20641
Medial Tibial Cartilage	9082	13412	18681	24546

The mesh density is determined by analyzing the variable of interest, in this case the contact pressure (CPRESS) on the cartilage, for change from the last mesh density. For these nonlinear models, mesh convergence was defined as a change in CPRESS of less than 10% at each node that was analyzed. When a simulation showed a less than 10% change in a given mesh body, the mesh body used in the prior simulation was chosen as the converged mesh and was used for all further simulations. The meshing process for this schema is constructed in a way that leaves several surface nodes in the same spatial location through different mesh densities. These nodes were located in each of the part files through the use of the MatLab script “findMatches.m” which can be found in G. MatLab Scripts. The nodes used were required to have a non-zero value for CPRESS for all iterations of the meshes. Ideally the nodes were also close to the zone that has the maximum CPRESS value for the body. Multiple nodes were analyzed on each mesh to ensure global convergence and not a local convergence and are referred by an ID number as the node number assigned by TetGen meshing varies with mesh density. When the mesh becomes too fine, the elements suffer from excessive distortion and the simulation

is unable to converge. A balance between number of elements and ability to converge is needed for a simulation to function correctly.

The mesh convergence study consisted of 4 models where the mesh density of the femoral cartilage, lateral tibial cartilage and medial tibial cartilage varied. Control of this meshing was difficult due to the number of variables involved, especially inconsistent segmentation and will be discussed more in 4.2.3.

2.3.11. Abaqus Simulation Validation

A converged mesh is important, but it is meaningless without a comparison to data from physical experiments. The mesh convergence process was conducted with an axial compressive load of 500 N with the tibia completely fixed and the femur fixed in the flexion-extension axis with no relative rotation. This closely resembles both the future simulations with the subject-specific load and boundary conditions and an in vitro study conducted on the TF joint by Seitz [35]. These load and boundary conditions were also used to understand the effects of the altered control parameters, especially the effect of viscous forces due to the global damping. The modeling scheme was considered validated if one of the subject-specific geometries had a CPRESS value that fell within one standard deviation of the reported mean pressure when subjected to the boundary conditions and loads discussed in [35].

3. RESULTS

3.1. Mesh Convergence

The results of the mesh convergence study are shown below with Figure 3-18 showing a representative contour plot of the variable of interest, CPRESS [MPa]. The following

figures and tables give the values of CPRESS at the nodes chosen for the mesh convergence study on each mesh body. The chosen mesh and the percent difference corresponding to the chosen mesh are highlighted in gray in each table.

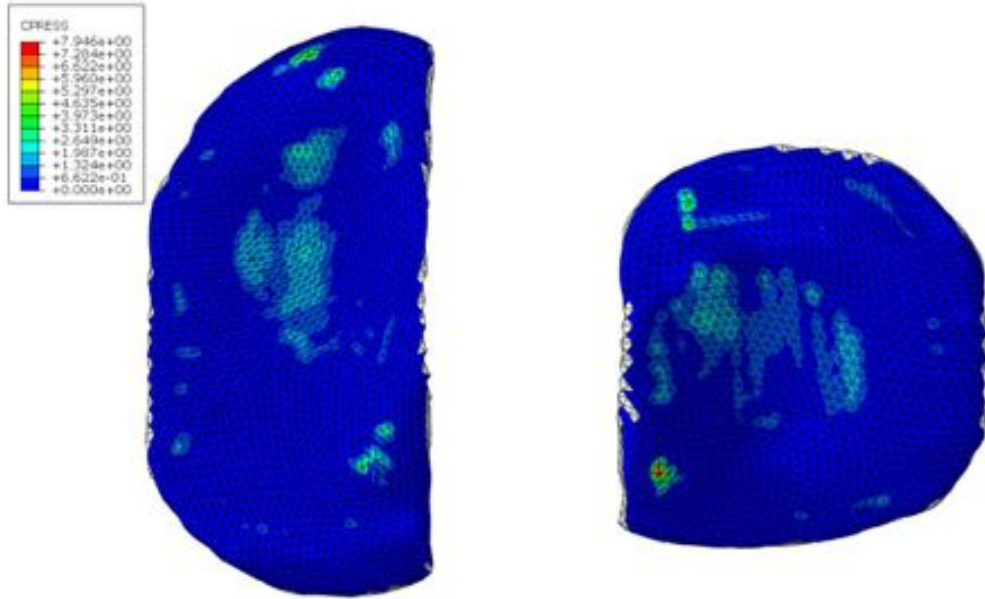


Figure 3-18. Contour Plot Of 2018Jan19-01 Geometry With CPRESS [Mpa] As The Variable Of Interest. Third Mesh Refinement.

Table 3-8. Results Of Mesh Convergence For Femoral Cartilage

Femoral Cartilage					
	Node ID	Mesh Refinement			
		1	2	3	4
CPRESS [MPa]	1	1.82	1.92	1.87	1.80
	2	1.75	1.59	1.60	1.62
	3	1.46	1.16	1.36	1.26
	4	0.40	0.38	0.47	0.51
% Difference	1		5.78%	-2.48%	-3.99%
	2		-9.02%	0.52%	0.94%
	3		-20.31%	16.36%	-7.25%
	4		-4.61%	21.20%	9.62%

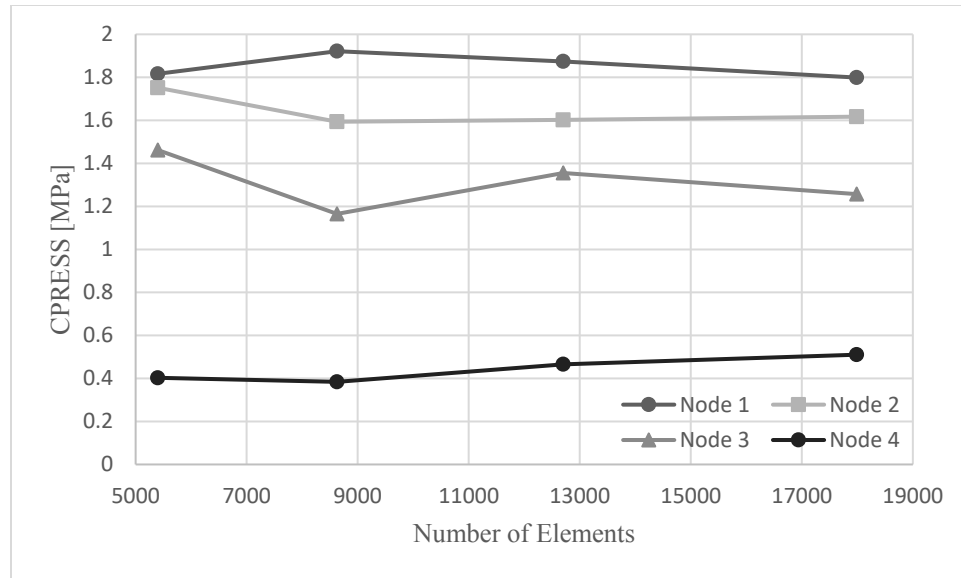


Figure 3-19. Visual Representation Of Variance In CPRESS [Mpa] For Femoral Cartilage Mesh Convergence. Number Of Elements In Mesh Body Vs. CPRESS At Chosen Nodes.

Table 3-9. Results Of Mesh Convergence For Lateral Tibial Cartilage

Lateral Tibial Cartilage					
	Node ID	Mesh Refinement			
		1	2	3	4
CPRESS [MPa]	1	1.46	1.30	1.36	1.36
	2	0.91	1.29	1.41	1.33
	3	1.48	1.18	1.19	1.65
% Difference	1		-10.87%	4.53%	-0.50%
	2		41.51%	9.64%	-5.57%
	3		-20.40%	1.02%	38.73%

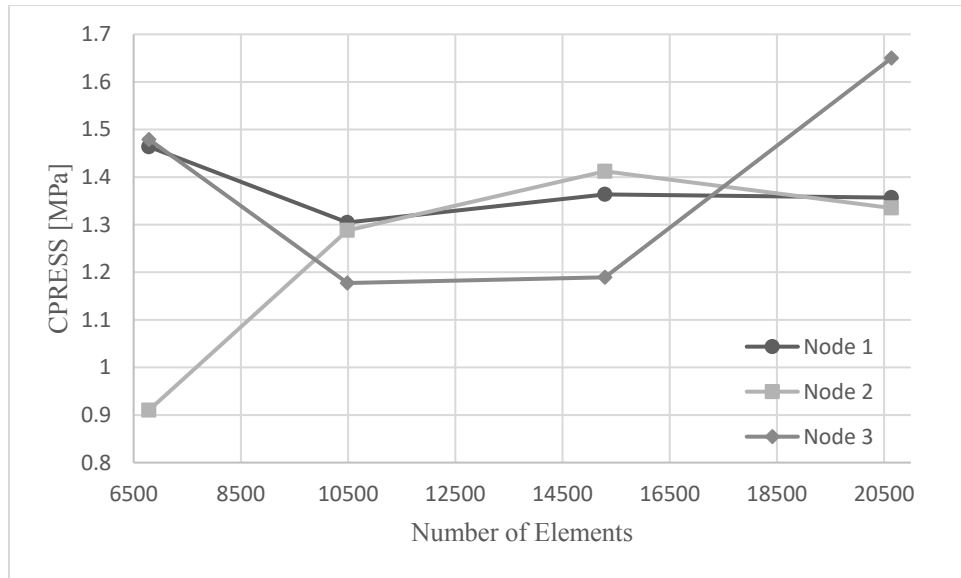


Figure 3-20. Visual Representation Of Variance In CPRESS [Mpa] For Lateral Tibial Cartilage Mesh Convergence. Number Of Elements In Mesh Body Vs. CPRESS At Chosen Nodes.

Table 3-10. Results Of Mesh Convergence For Medial Tibial Cartilage

Medial Tibial Cartilage					
	Node ID	Mesh Refinement			
		1	2	3	4
CPRESS [MPa]	1	1.12	1.15	1.20	1.15
	2	1.62	1.54	1.70	1.87
	3	0.18	0.48	0.90	0.90
	4	0.83	0.85	0.87	0.90
% Difference	1		2.63%	4.17%	-3.74%
	2		-4.56%	10.32%	9.75%
	3		164.23%	85.42%	0.37%
	4		2.17%	1.77%	3.68%

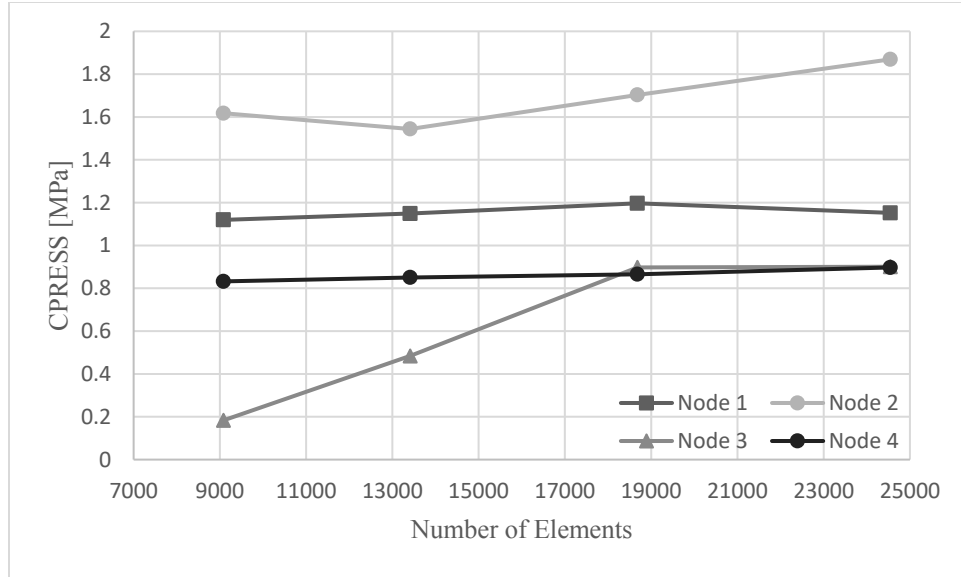


Figure 3-21. Visual Representation Of Variance In CPRESS [Mpa] For Femoral Cartilage Mesh Convergence. Number Of Elements In Mesh Body Vs. CPRESS At Chosen Nodes.

3.2. Validation Study

Table 3-11 and Table 3-12 below compare the FE simulation results to the experimental results from Seitz [35]. A cell with a CPRESS value highlighted in green indicates the value of CPRESS falls within one standard deviation of the experimental values. A cell containing the energy ratio is highlighted in green if the ratio of static dissipation energy to total internal energy is within the range of confidence per Abaqus FEA recommendations. Figure 3-22 is the magnitude of ALLIE (internal strain energy) and ALLSD (static dissipation energy from viscous forces) for the 500 N compressive load simulation. The x-axis ranges from 0 – 1 second, since this simulation is assumed pseudo static each second represents a full step. This plot shows that for all time ALLIE is greater than ALLSD, further confirming that the damping factor is not affecting physiological accuracy of this simulation. This result uses an automatically calculated damping factor meant to optimize the impact of damping.

Table 3-11. Results For Validation Study With 500 N Compressive Force Applied

[MPa]	Seitz 500 N Load			FEA	
	Mean - SD	Mean	Mean + SD	CPRESS	Energy Ratio [%]
Medial Tibial Cartilage	1.18	1.78	2.38	1.93	3.04%
Lateral Tibial Cartilage	0.97	1.79	2.61	1.88	

Table 3-12. Results For Validation Study With 1000 N Compressive Force Applied

[MPa]	Seitz 1000 N Load			FEA	
	Mean - SD	Mean	Mean + SD	CPRESS	Energy Ratio [%]
Medial Tibial Cartilage	1.95	2.82	3.69	3.41	1.56%
Lateral Tibial Cartilage	1.72	3.04	4.36	3.07	

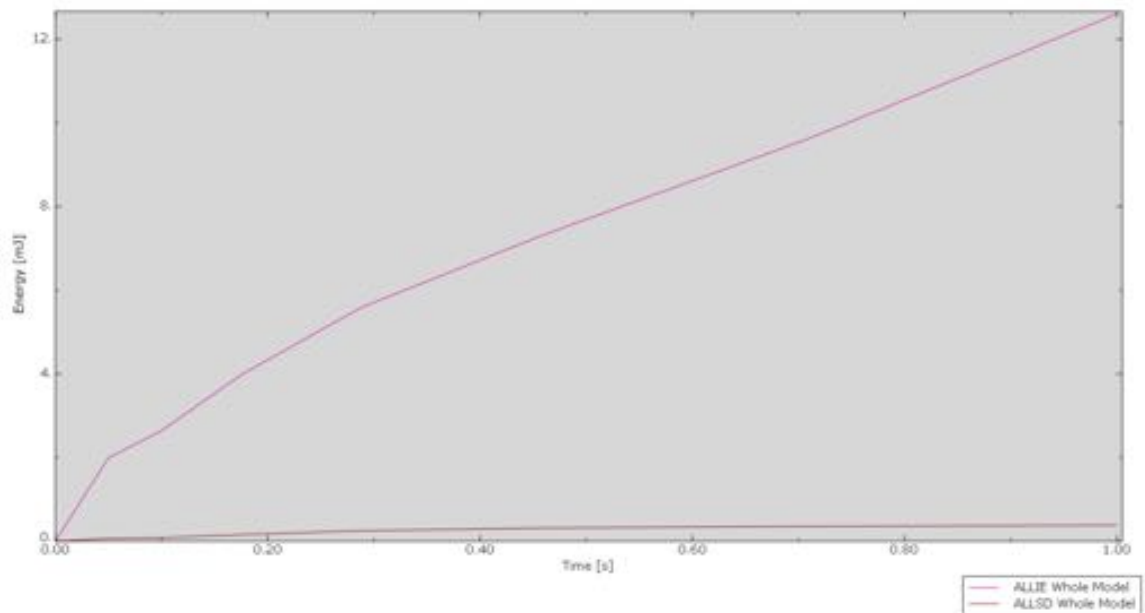


Figure 3-22. Plot Of Total Internal Energy (ALLIE) And Static Dissipation Energy (ALLSD) Vs Time For A One-Step Simulation With Boundary Conditions Matching The Experimental Study At 500 N Of Compressive Force. Automatic Stabilization Scheme Used.

3.3. Joint Contact Load Simulations

The contour plots of the contact pressure (CPRESS) [MPa] for the medial and lateral tibial cartilage mesh bodies for all nine simulations, three for each participant, are shown in Figures 3-22 through 3-30. The accompanying tables (3-13 through 3-15) give the numerical value of max CPRESS based off the methods discussed in 2.3.9 for each exercise and participant. If a simulation was unable to converge to completion but was within approximately 10% of convergence in the loading step, the value for CPRESS was linearly extrapolated for the remainder of the step, an asterisk denotes this.

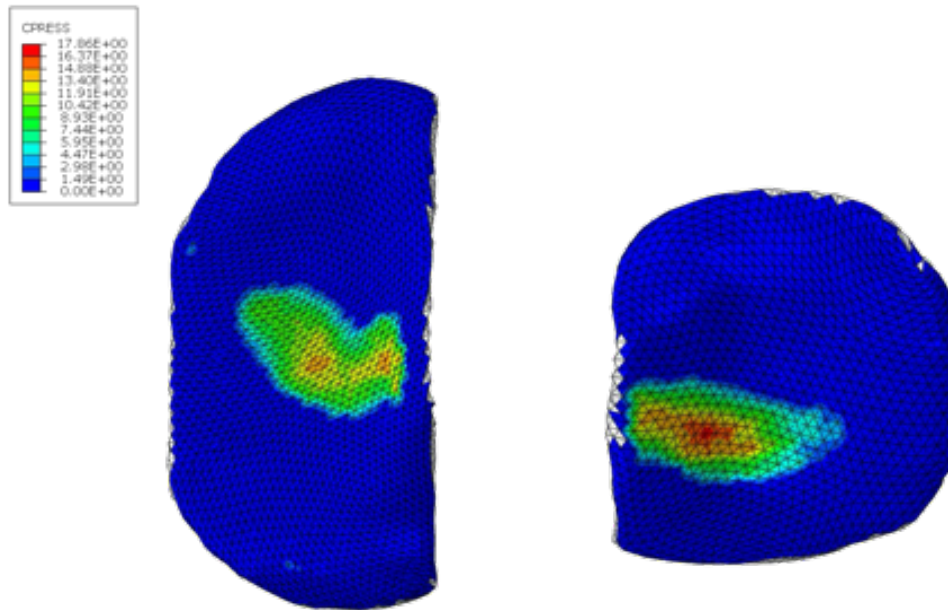


Figure 3-23. Contour Plot With CPRESS Plotted For Gait Results Of 2018Jan19-01 (Control)

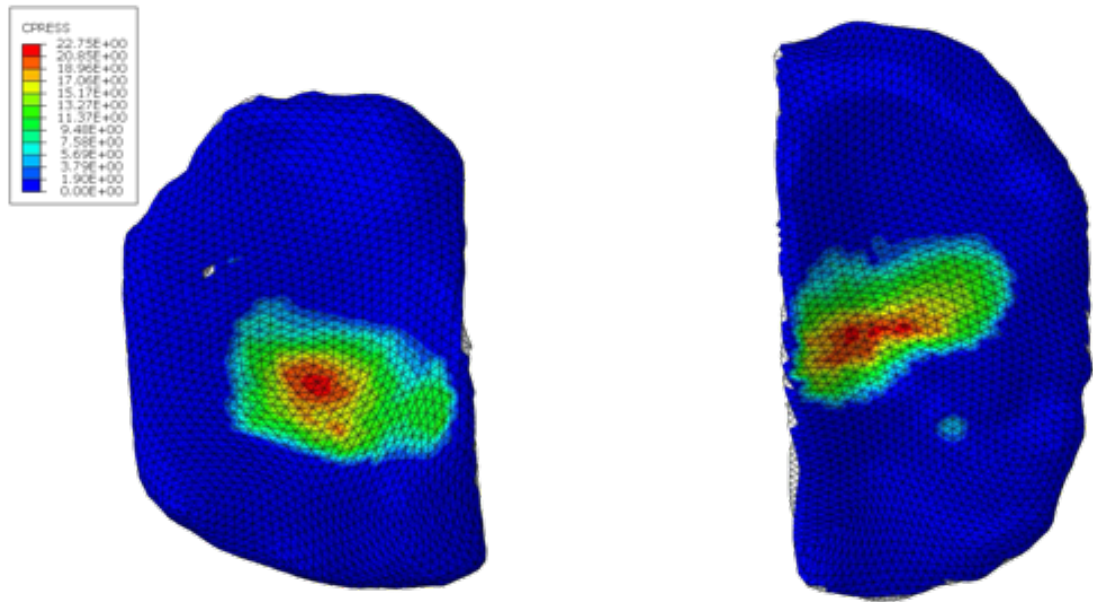


Figure 3-24. Contour Plot With CPRESS Plotted For Gait Results Of 2016Nov10-01 (Amputee)

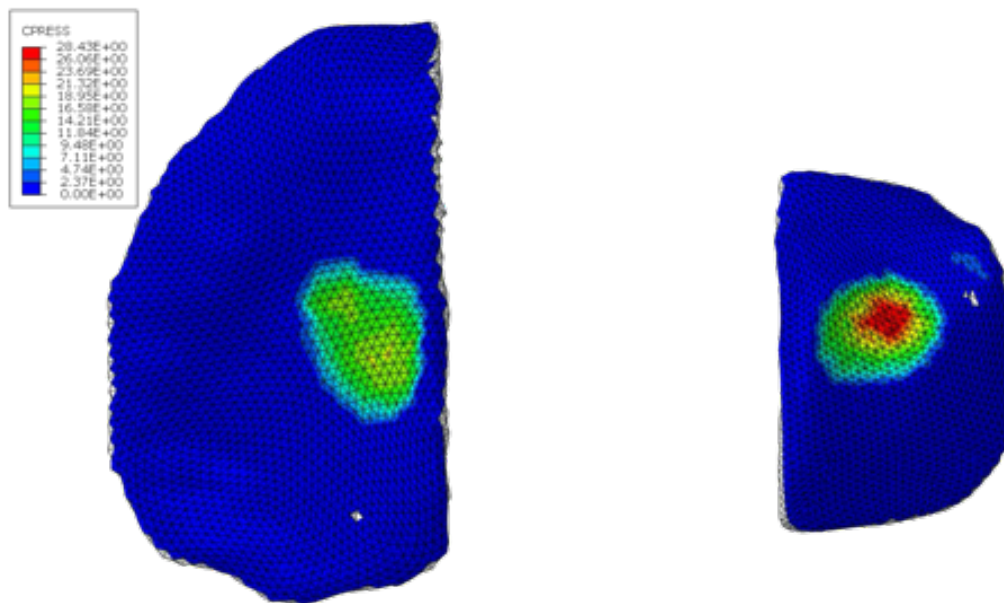


Figure 3-25. Contour Plot With CPRESS Plotted For Gait Results Of 2016Nov14-02 (Amputee)

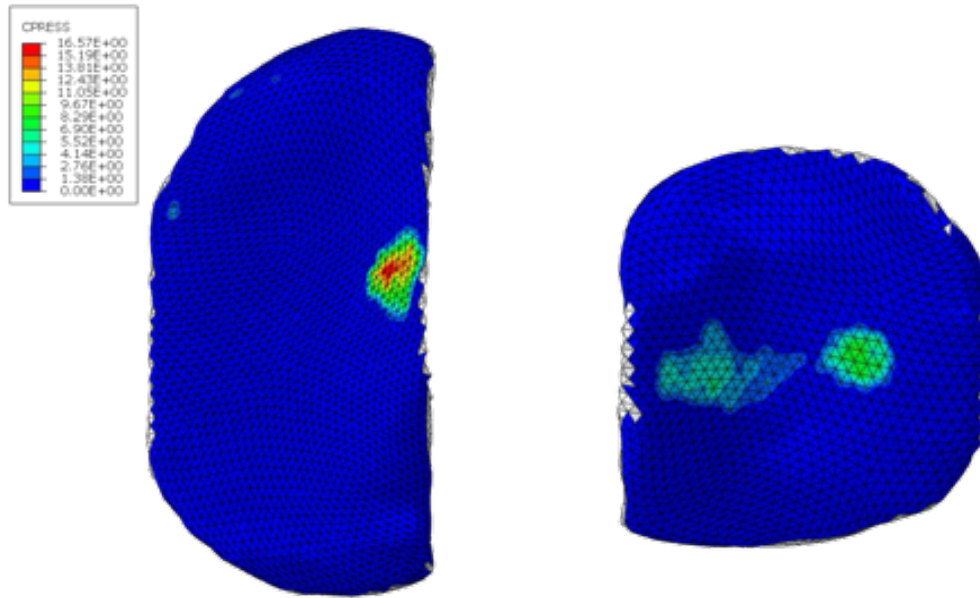


Figure 3-26. Contour Plot With CPRESS Plotted For Cycling Results Of 2018Jan19-01 (Control)

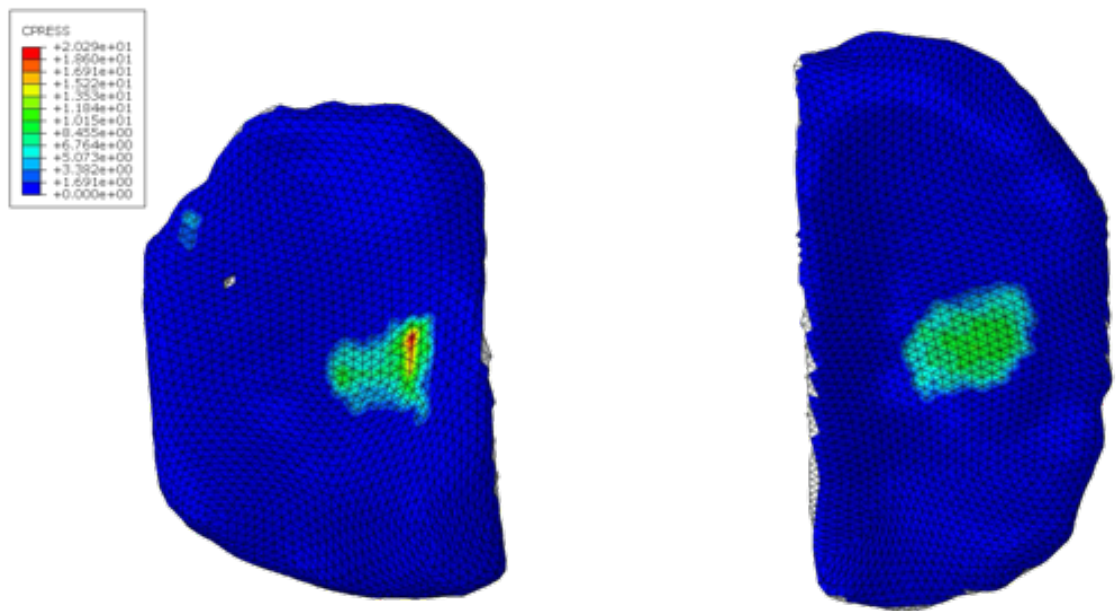


Figure 3-27. Contour Plot With CPRESS Plotted For Cycling Results Of 2016Nov10-01 (Amputee)

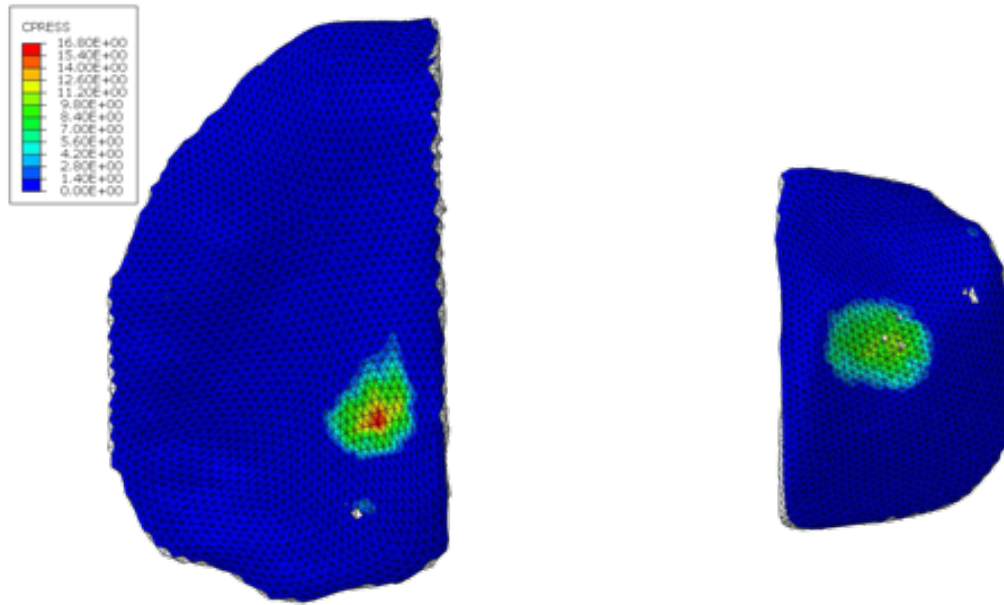


Figure 3-28. Contour Plot With CPRESS Plotted For Cycling Results Of 2016Nov14-02 (Amputee)

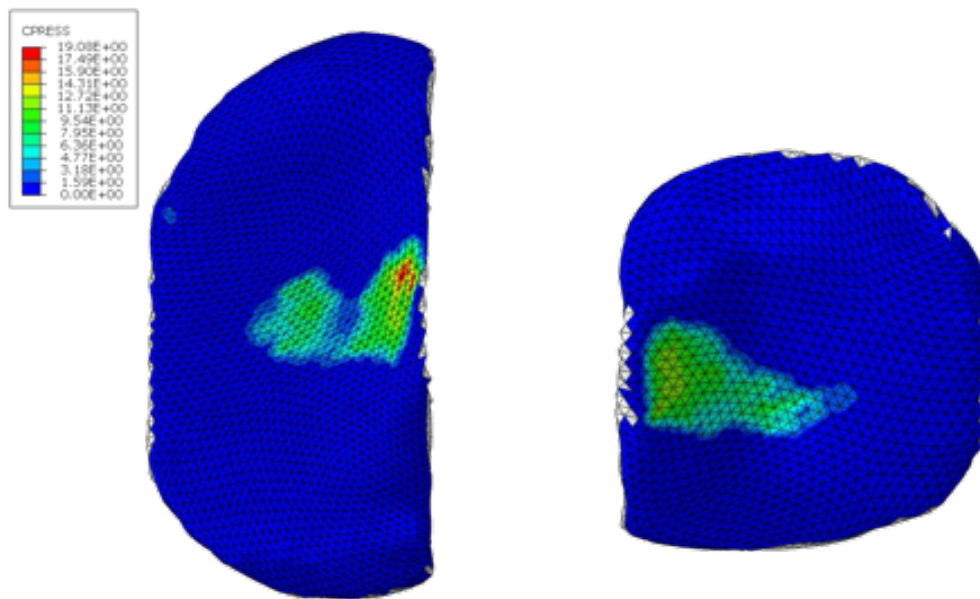


Figure 3-29. Contour Plot With CPRESS Plotted For Elliptical Results Of 2018Jan19-01 (Control)

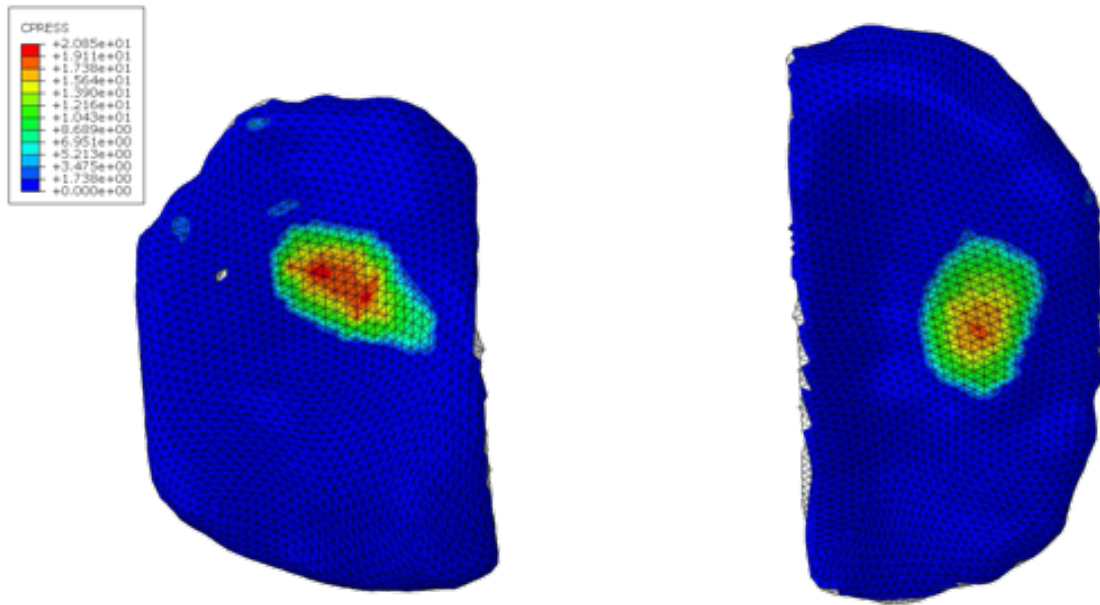


Figure 3-30. Contour Plot With CPRESS Plotted For Elliptical Results Of 2016Nov10-01 (Amputee)

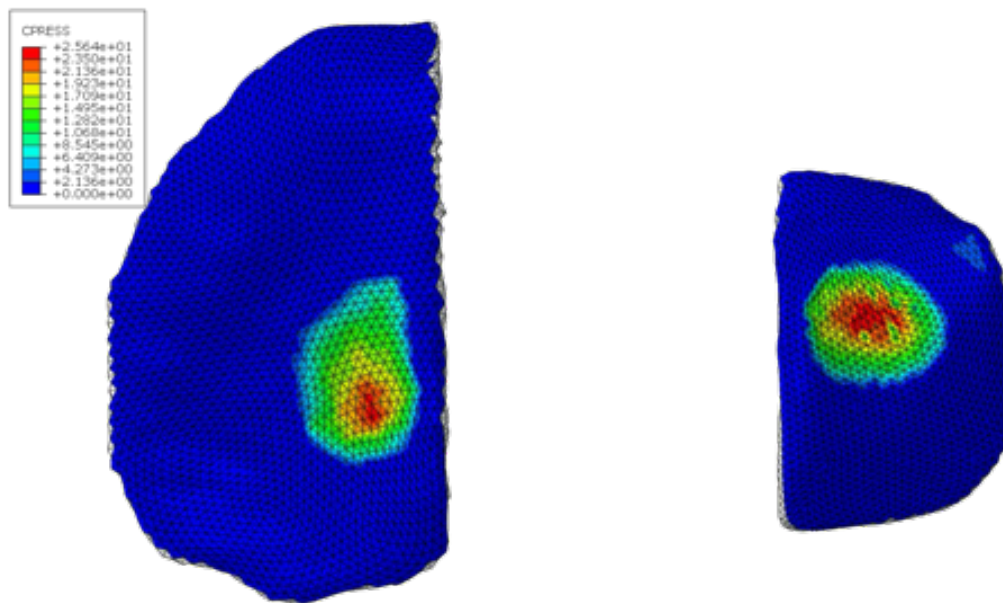


Figure 3-31. Contour Plot With CPRESS Plotted For Elliptical Results Of 2016Nov14-02 (Amputee)

Table 3-13. Joint Contact Load Simulation Results For Gait Exercise

	CPRESS [MPa]		CPRESS normalized by bodyweight [MPa/N]	
	Lateral Tibial	Medial Tibial	Lateral Tibial	Medial Tibial
2018Jan19-01 (Control)	15.86	13.68	0.020	0.018
2016Nov10-01 (Amputee)	21.44	20.81	0.029	0.028
2016Nov14-02 (Amputee)*	29.23	19.73	0.036	0.024

Table 3-14. Joint Contact Load Simulation Results For Stationary Cycling Exercise

	CPRESS [MPa]		CPRESS normalized by bodyweight [MPa/N]	
	Lateral Tibial	Medial Tibial	Lateral Tibial	Medial Tibial
2018Jan19-01 (Control)	7.52	14.16	0.010	0.018
2016Nov10-01 (Amputee)	10.32	13.33	0.014	0.018
2016Nov14-02 (Amputee)*	10.81	13.96	0.013	0.017

Table 3-15. Joint Contact Load Simulation Results For Elliptical Training Exercise

	CPRESS [MPa]		CPRESS normalized by bodyweight [MPa/N]	
	Lateral Tibial	Medial Tibial	Lateral Tibial	Medial Tibial
2018Jan19-01 (Control)	11.09	15.17	0.014	0.020
2016Nov10-01 (Amputee)	18.84	17.40	0.026	0.024
2016Nov14-02 (Amputee)*	23.89	21.78	0.029	0.026

The maximum contact pressure has been reported as the raw value along with a value normalized by bodyweight [MPa/N]. The raw value is useful for comparing the medial and tibial compartments to each other within the same participant and for future studies to compare results to while the normalized value is useful for comparing each participant. Normalizing based off bodyweight removes the confounding variable of non-uniform

bodyweight of the participants to alleviate false patterns forming from the raw contact pressure values.

With the current results, it is not possible to conduct a statistical analysis to compare the results from a single control participant with the two amputee participants and detect statistically significant differences. Qualitatively, the body mass-normalized contact pressure experienced by the articular cartilage of the amputee simulations appears to be larger in magnitude than the contact pressure in the control model in both the medial and lateral tibial cartilage compartments. There is one exception with the medial tibial cartilage contact pressure in cycling where the control participant has the same body weight normalized contact pressure as one of the amputee participants. Based on the participants analyzed in this study, the results show gait and elliptical exercises have larger contact pressure than cycling in both compartments. Gait and elliptical comparisons are not as clear with gait having larger contact pressure than elliptical for some participants in some compartments but there is no distinct pattern. The difference in magnitude of contact pressure between gait and elliptical for each compartment and for each participant is minimal. For all participants in cycling, the magnitude of contact pressure is larger in the medial compartment than the lateral compartment of the tibial cartilage.

Visually, the contact pressure for the gait exercise maintains a magnitude closer to the maximum value reported for a larger cartilage area than the cycling and elliptical results. Simply put, the cycling and elliptical simulations tend towards a smaller area of high contact pressure while the gait simulations show a large area of high contact pressure (compare the colored regions in Figures 3-22 to 3-30).

This visual can also be quantified through a model parameter called CAREA or the total surface area of a contact surface that is actually experiencing contact. The results are shown in Table 3-16 below.

Table 3-16. Total Surface Area [Mm²] (CAREA) Of Tibial Cartilage Compartments In Contact With Femoral Cartilage

CAREA (mm ²)	Gait		Cycling		Elliptical	
	Lateral Tibial	Medial Tibial	Lateral Tibial	Medial Tibial	Lateral Tibial	Medial Tibial
2018Jan19-01 (Control)	188.6	184.9	120.3	42.2	152.4	149.5
2016Nov10-01 (Amputee)	212.7	203.0	65.2	80.5	104.9	103.9
2016Nov14-02 (Amputee)	70.3	100.8	49.1	42.7	93.3	115.3

Qualitatively, the contact area for gait and elliptical exercise is larger than cycling for both compartments. For gait, two of the participants have more area in contact than elliptical but 2016Nov14-02 has elliptical contact area slightly larger than gait.

Comparing lateral to tibial compartments, the total area in contact is similar. The largest difference in contact area magnitude is for 2016Nov14-02 in gait with the medial compartment contact patch having 30.5 mm² more surface area than the lateral compartment contact patch, roughly 30% different.

4. DISCUSSION

4.1. Results

4.1.1. Convergence Analysis

The convergence analysis demonstrated that the variable of interest converged within 10%, which is better than previous studies of whole knee FE models by the Cal Poly

HMB. This schema succeeds in a more robust and consistent simulation than previously created by the Cal Poly HMB based off the convergence study results.

4.1.2. Validation Study

The results from the validation study with a 500 N compressive force and a 1000 N compressive force applied to the femur show agreement with the experimental study performed by Seitz in [35]. In both load cases, both the medial and lateral tibial cartilage contact pressures fall within one standard deviation of the mean values reported by Seitz. Additionally, the static dissipation energy to total internal energy ratio for both load cases was below the recommended maximum when a global damping ratio was applied for stabilization. The subject-specific finite element models created for this study have been successfully validated against experimental data.

4.1.3. Joint Contact Load Studies

The core objective of this thesis was to simulate the effects of gait, stationary cycling and elliptical training exercises on the articular cartilage in the tibiofemoral joint. Specifically, a comparison was performed between control participants and transtibial amputee participants when examining the magnitude of contact pressure exerted on the tibial articular cartilage. This was done using subject-specific finite element models composed of subject-specific geometry gathered from MRI scans and subject-specific joint contact loads gathered from motion analysis experiments. To do this, a consistent schema capable of producing FE models robust enough to converge with large deformations was developed. The goals outlined were partially met.

This schema was able to produce converged models when forces were applied that were more than three times larger in magnitude than previous Cal Poly HMB lab studies,

again demonstrating more robust models than previous whole knee models developed in the HMB lab. Out of the three geometries created, there were nine simulations attempted, three simulations for each subject-specific tibiofemoral joint geometry that correspond to the three exercises performed in the motion analysis experiments. The cartilage contact pressures determined by these models are the most physiologically accurate measurement made thus far by Cal Poly HMB lab members and can stand next to the large body of work in this field.

The body of research that specifically looks at FE models of the tibiofemoral joint articular contact pressure is limited, the number of studies that look at transtibial amputees is even more limited. This makes it difficult to directly compare the results of the subject-specific maintenance exercise simulations. The closest results are from Lane in [25]. Table 4-17 and Table 4-18 below reports the values from Lane in comparison with the values from these simulations for both gait and cycling exercises.

Table 4-17. Contact Pressure Results For Gait Exercise. Average Values From Lane And Subject-Specific Values From This Thesis.

CPRESS Normalized by bodyweight [MPa/N]		Lane	Stearns		
		Average CPRESS	2018Jan19-01 (C)	2016Nov10-01 (A)	2016Nov14-02 (A)
Control	Medial	0.0157	0.0176	-	-
	Lateral	0.00896	0.0205	-	-
Amputee	Medial	0.0163	-	0.0284	0.0240
	Lateral	0.00883	-	0.0292	0.0355

Table 4-18 Contact Pressure Results For Cycling Exercise. Average Values From Lane And Subject-Specific Values From This Thesis.

CPRESS Normalized by bodyweight [MPa/N]		Lane	Stearns		
		Average CPRESS	2018Jan19-01 (C)	2016Nov10-01 (A)	2016Nov14-02 (A)
Control	Medial	0.0119	0.0183	-	-
	Lateral	0.00391	0.00970	-	-
Amputee	Medial	0.0123	-	0.0182	0.0170
	Lateral	0.00363	-	0.0141	0.0131

For gait, the one control simulation from this thesis has a cartilage contact pressure value approximately 12% larger than the average from Lane [25] in the medial compartment and 128% larger in the lateral compartment. The amputee simulations have contact pressure values 74% and 47% larger in the medial compartment and 230% and 302% larger in the lateral compartment.

For cycling, the one control simulation from this thesis has a contact pressure value approximately 53% larger than the average from Lane [25] in the medial compartment and 148% larger in the lateral compartment. The amputee simulations have contact pressure values 47% and 38% larger in the medial compartment and 287% and 262% larger in the lateral compartment.

These results do make physiological sense as Lane applied joint reaction loads in his simulation rather than joint contact loads. On average, joint contact loads are 200-300% larger in magnitude than joint reaction forces. For this thesis, the varus-valgus and internal-external rotation joint contact moments were neglected as the current methodology from Fernandez in [32] only utilizes the forces. With the applied varus-valgus moment from Lane, the contact pressure is shifted to the medial compartment and

results in much larger medial contact pressure than lateral contact pressure. This thesis found there to be very slight differences in the contact pressure when comparing the two tibial compartments in the same simulation. The difference in these two results is very likely the result of neglecting the varus-valgus moment. Because muscle forces are not included in the loading case for this thesis, the femur stays axially centered above the tibia. A further study should see if including the varus-valgus moment alters the distribution in contact pressure between the tibial compartments.

In gait for this thesis, the lateral compartment had a larger contact pressure value than the medial and in cycling the medial compartment had a larger contact pressure than the lateral for each simulation.

There is a small body of work that has also created subject-specific FE models of the TF joint and applied subject-specific boundary conditions derived from motion analysis experiments. One study from Klodowski in 2015 [57] created one such model for gait for a single subject. This paper details a model that uses force and flexion-extension angle curves from motion analysis data as boundary conditions for the FE model. However, this study only utilized compressive force on the model most similar to this thesis (Model A in the paper). Once normalized over bodyweight, the contact pressure reported in this thesis for the medial compartment of the control participant was very similar to that in [57], this thesis had a magnitude approximately 15% larger. The agreement in the lateral compartment was not as good with about a 300% difference. This could be attributed to a difference in boundary conditions with the femur in [57] having less ability to rotate about the varus-valgus axis.

Another study from the Cal Poly HMB lab from Orekhov also supports the findings of this thesis. In [34] Orekhov found that the maximum compressive joint resultant force in the tibiofemoral joint was highest in gait and lowest in cycling, with elliptical slightly lower than gait. This aligns with the results from this thesis showing that the contact pressure in gait is largest with cycling being the smallest and elliptical slightly lower than gait.

Comparing the contact pressure of the tibial cartilage in this thesis with other papers that are also tibiofemoral joint FE models such as [35], [45], [54] that address the effects of meniscectomies and body weight on joint contact pressure also show the results found in this thesis are reasonable. The applied compressive forces in these papers are around three to four times less than the compressive forces in this thesis, the resulting contact pressures are also three to four times less than the contact pressures reported in these papers. This makes physical sense as the relationship between axial force and contact pressure is linear ($F/A = P$).

This thesis succeeds in generating fully subject-specific finite element models that can produce meaningful, physiological data. Due to the implementation of joint contact forces with subject-specific load data, these models represent the most accurate whole knee finite element models to be developed by the HMB lab and the only subject-specific whole knee models using subject-specific loads for controls and transtibial amputees ever created.

4.2. Limitations and Recommendations

This schema was developed with the goal of a low-cost semi-automated process for gathering subject-specific data and creating a functional FE model of the TF joint. Based

off the results and from many trials throughout the project, it is evident that this schema is a step in the right direction but is not a flawless methodology. The limitations discussed will be partnered with suggested future work.

4.2.1. MRI Images and Segmentation

A local hospital was chosen for the MRI scans of the TF joint for all participants. This choice was made with practicality in mind for the subjects who volunteered for the experiments and allowed the researchers to communicate with the medical professionals who performed the scans. The MRI machine available at the medical facility has a field strength of 1.5T, which is generally accepted as sufficient for diagnoses involving large bones in the torso and legs. However, a machine capable of producing a 3.0T field is more optimal for segmentation because the scans produce more clear and vivid images. See Figure 4-32 and Figure 4-33 below to compare, with Figure 4-32 using a 1.5T machine and Figure 4-33 using a 3.0T machine.



Figure 4-32. MRI Scan Of TF Joint From The Sagittal Plane From A 1.5T Machine

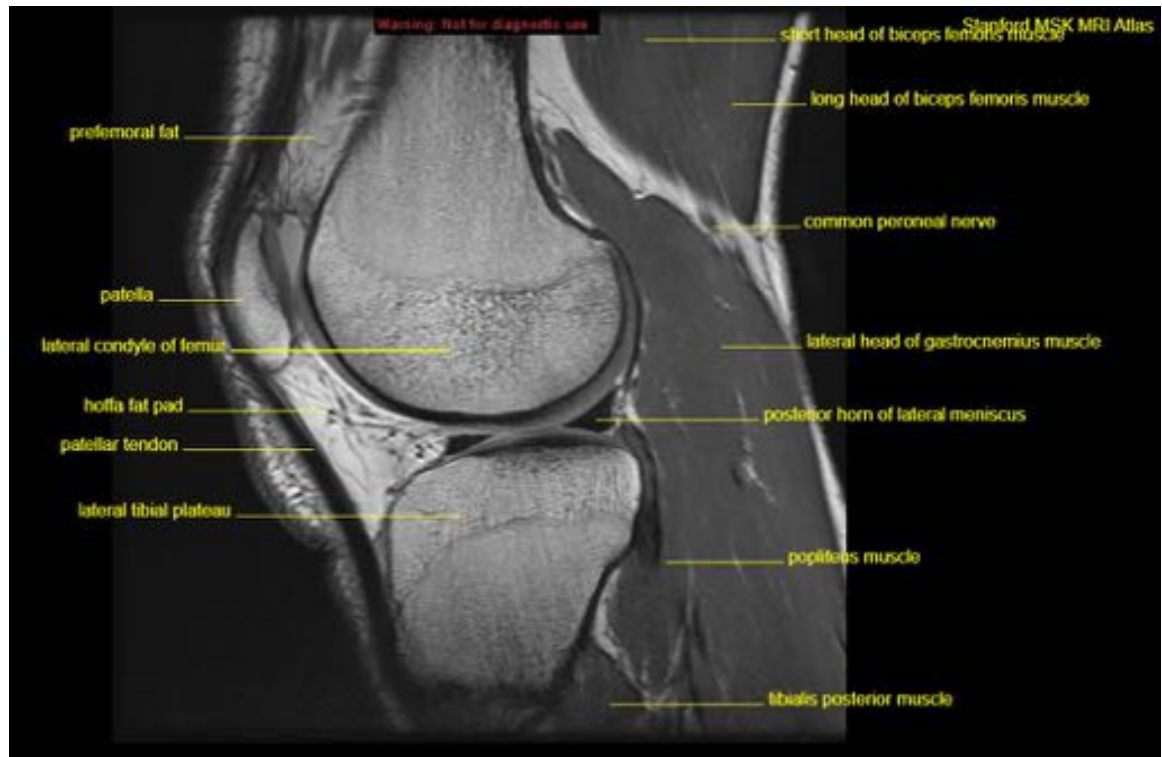


Figure 4-33. MRI Scan Of TF Joint From The Sagittal Plane From A 3.0T Machine [58]

The difference in clarity is pronounced and any future MRI scans should be obtained from a machine that uses a 3.0T field. This change would help immensely with issues that developed from the segmentation process. It was very difficult to determine the boundaries for the bodies used in the model and impossible to accurately segment ligaments.

The segmentation process relied heavily on user judgement for the boundary of each body. This raises problems when the user is not a trained medical professional. While working on a project where anatomical accuracy is of vital importance, the individual doing the segmentation should have some level of professional training and guidance. One model (Subject ID: 2019Jan19-01) was validated by a local orthopedic surgeon for physiological accuracy. A recommendation for further work on this project is to recruit a single researcher who has familiarity with human knee anatomy to fully segment all

bodies and to have all these segmentations validated by a local orthopedic surgeon. Based off conversations with local radiologists, there will always be some level of inconsistency from individual to individual when reading MRI scans. Using a single researcher to segment all bodies will lead to consistent judgement calls across all models with the added benefit of improvement over time with reading the images.

Additionally, manual segmentation was used in ITK-SNAP which lead to user error due to the very slight changes in contrast that denote a change in tissue density. Automated segmentation does not suffer from bias like a manual segmentation. Once the parameters have been defined for a single knee, they may be applied to all future knees to ensure consistency if not perfect accuracy. The automated segmentation for open-source software is not well documented so the attempts made during this project were not successful. This could also be due to a lack of contrast between tissue densities that are a product of the 1.5T scans. A potential replacement for open-source segmentation is the commercial software Materialise Mimics, which supports automated and manual segmentation as well as geometry smoothing.

4.2.2. Smoothing

With a schema emphasizing open-source solutions where possible, the geometry smoothing performed with MeshLab proved the most successful part. While documentation of the filters used was lacking and a concise user guide is not available, nearly all filters reference an article that describes the algorithms used. The filters are not perfectly stable and especially with isoparameterization it is difficult to reproduce similar results even with the same starting geometry. Results were close enough to be computationally irrelevant provided no errors occurred through the smoothing process.

This software could be alongside a commercial segmentation software with functional results, but if the commercial software possesses its own smoothing algorithms a comparison could be made between the two.

There were several of the soft tissue bodies that had volumes change by more than the desired 5%. This is likely due to segmentation errors, as the bodies in question had many vertices removed with the filter intended to remove unconnected regions while others that had less than 5% change had very few. This issue with smoothing would be eliminated by more accurate segmentation. The most obvious method for more accurate segmentation is to utilize the automated segmentation in ITK-SNAP or Mimics with a well-documented set of parameters for distinguishing tissues so error by the user is minimized. This may not be possible with the MRI machine used for this study, so an alternative method is to recruit an individual with a primary objective of creating the most accurate and smooth segmentation of the tissues while verifying each body with an expert in the field.

4.2.3. Meshing

The meshing process is arguably the most important step for a finite element model. Without well-defined elements that are free of errors, convergence of a model is impossible as an inverse of the stiffness matrix must be calculated to solve the simulation. TetGen is a promising open-source solution for any simulations where tetrahedral elements are desired. It allows for fully automated meshing when provided with a closed surface and can output a file compatible with many FE solvers. However, TetGen does not provide a simple method of creating surface meshes such as those used for the bones and the computational meshes it does output do not always reflect the expected results.

The most important flaw is that TetGen can produce elements that have 0 volume by the standards of Abaqus FEA, as well as a number of distorted elements on the bodies used in this study. The soft bodies are highly irregular in shape and likely require refinement in specific areas to produce a mesh that does not have an excessive number of elements. There are potential confounding variables, the imported mesh from MeshLab could have triangular facets that do not allow for a tetrahedral volume mesh to generate correctly. All elements that had 0 volume or were highly distorted when the bodies were imported into Abaqus FEA were removed with a custom MatLab script included in G. MatLab Scripts. This is not an ideal solution as it introduced localized stress concentrations that could make results inaccurate in the areas of interest.

Most importantly, hexahedral meshing is not included within TetGen. While studies have shown that quadratic tetrahedral elements can perform just as well as linear hexahedral elements [52], it remains that hexahedral elements are more robust when large distortions occur. The main convergence issue experienced was negative eigenvalues, this error occurs when an element distorts to the point that it self-intersects and has zero volume. Tetrahedral elements suffer from this more often than hexahedral elements due to the difference in the number of sides, especially when subjected to large forces normal to the surface of the element such as in contact. Since there are no open-source hexahedral meshing solutions with sufficient documentation available, it is highly recommended that a commercial meshing software that allows for meshing with quadratic hexahedral meshing is used in future studies. Two options include TrueGrid (XYZ Scientific Applications, Pleasant Hill, CA) and Hypermesh (Altair Engineering, Troy, MI). Previous HMB works including [22]–[24] used TrueGrid for the FE mesh

generation. Both softwares use a similar approach of creating a domain of cubes that is projected on the bodies with several steps to create a best fit mesh of the curves. This is a manual process and can introduce error, especially when the user is not familiar with the methodology. TrueGrid does have a goal of automating the hexahedral meshing process but it is unclear when and if this goal will be attained. By eschewing an automated process, the overall time invested into a single subject-specific FE model increases, however it has been shown by this methodology that automated tetrahedral meshing has its own limitations with a lack of control over the final mesh. A specific paper to reference a comprehensive meshing scheme is Besier in [16]. Besier uses a commercial software, Geomagic Wrap (3D Systems, Research Triangle Park, North Carolina), to create NURBS surfaces from point cloud data. This eliminates the need to mesh bodies outside of Abaqus as the NURBS surface may be exported as the Abaqus-compatible file format IGES. This will eliminate the confounding factor zero-volume elements being present in Abaqus when TetGen reports none. While the automated meshing of TetGen is convenient and has a low barrier to use, a robust mesh is more important than some time savings in the meshing process.

In addition, using two separate software to obtain a desired average edge length of a computational mesh is difficult, especially when MeshLab does not have this capability easily available. This led to the parallel bodies across each knee joint having significantly different average element sizes despite careful control over the number of surface faces. A meshing program that allows the user to set the desired average edge length in various regions would allow for much more robust meshes.

4.2.4. Assembly Components and Material Properties

Various issues arising from segmentation have been discussed, a final limitation concerning the modeling of ligaments also is derived from segmentation. The TF joint has an extremely complex set of ligaments that allow for stability during activity, they are attached both to bones and to the menisci. This system in this study has been approximated as four non-linear axial springs, based off prior work. Ideally, ligaments would be modeled as nonlinear anisotropic bodies that interact with all components of the TF joint. A large portion of ligament stiffness comes from the friction between the ligament bundles as they wrap around each other and elongate when under tension and from viscous forces exerted by the synovial fluid surrounding the ligament bundles during elongation. The MRI scans did not produce images that allowed for continuum bodies to be generated for the ligaments. In addition, very little research exists to justify the approximation of these ligaments as simple springs, as quantifying the material properties of ligaments in vitro is vastly different than the behavior in vivo.

The meniscal horns were also neglected as the previous method of using nonlinear springs to act as the attachment horns led to excessive motion of the menisci as they slid anterior to the tibial plateau relative to the cartilage they rest on. With the method of tying the menisci to the cartilage, this issue is eliminated, but it does not reflect the physiological behavior of the menisci. This could lead to inaccurate placement of the femur after it has undergone rotation. The menisci are meant to ensure the femur stays located over the tibial plateau, which is maintained with the methods used to fix the menisci in this paper however it may be too severe as there is minimal relative translation

between the femur and the tibia in the simulation. It is unknown if this is physiologically accurate and future work should research studies that discuss this behavior.

Additionally, both the meniscus and cartilage are modeled as linear elastic materials. This is a common practice for quasi-static analyses, but it does not reflect the meniscus behavior in vivo. The meniscus provides stability to the TF joint during activity through its anisotropic properties distributing an unstable compressive force into hoop stress [59]. In the simulations performed with this schema, the meniscus takes almost no load due to the frictionless contact forcing the knee to slide off the nearly incompressible meniscus. Because this study is more interested in cartilage contact pressure and because the role of the meniscus is to redirect the femur contact to the central portion of the tibial cartilage, this behavior is acceptable but not ideal.

4.2.5. Load and Boundary Conditions

The aim of the research conducted by the Cal Poly Human motion biomechanics lab is to understand the cause of early onset osteoarthritis in the TF joint experienced by transtibial (TT) amputees. The hypothesis was presented that TT amputees have a larger magnitude of contact pressure on the articular cartilage which degrades the tissue. The schema in this paper uses the load conditions derived from the point at which maximum compressive force occurs. This force tends to be the largest in magnitude when compared to the anterior-posterior shear and medial-lateral shear which leads to an assumption that the largest compressive force likely means large contact pressure. However, one time point during a cycle of gait, cycling, and elliptical training does not represent daily activity that is a normal cause for osteoarthritis. Further studies could explore the entire

cycle through multiple steps in Abaqus FEA to better represent the full loading behavior of the cartilage in daily activity.

This study also neglected muscle forces that affect the TF joint. Previous work shows that muscle forces can be a large contributor to joint contact forces [60]. The effect of applying muscles forces to the FE model is unknown. The contact forces used in this thesis were determined from a multi-body simulation with muscle actuators and should be fairly accurate. Since the contact forces and the tibial constraints are specified, it is possible that adding muscle forces directly to the tibia would only change the reaction forces required to maintain the zero displacement (translational and rotational) constraints on the tibia. It is unclear what effect adding the muscle forces to the femur would have. Additionally, the effects of including the patellofemoral joint contact force along with a patellar force at the insertion point on the anterior tibia are unknown. Future studies should attempt to include these forces and examine their impact.

More importantly, the varus-valgus and internal-external rotation moments were neglected. This data was not available at the time this study was conducted, however it could explain why there was not a large difference between the medial and lateral compartments for exercise with previous work from Lane [25] showing the opposite.

4.2.6. Simulation Controls and Damping

The effect of viscous damping on the simulations utilizing experimental knee angles and joint contact loads is much higher than the recommended levels. As shown earlier, the automatic stabilization scheme produces a simulation that minimizes the impact of the damping factor by keeping the static dissipation energy much lower than the total internal energy in the model when applied to the validation study. The experimental studies were

unable to converge using the automatic stabilization scheme and required a user-defined constant factor to converge. An exploratory study was performed where automatic stabilization was used during the rotation step while a constant damping factor was used in the loading step where Figure 4-34 below shows both energies over time. It is clear the constant damping factor causes the static dissipation energy to dominate the simulation. A study focusing on a combination of automated and constant damping could make this modeling scheme viable for completely subject-specific simulations utilizing subject-specific geometry and subject-specific joint contact loads.

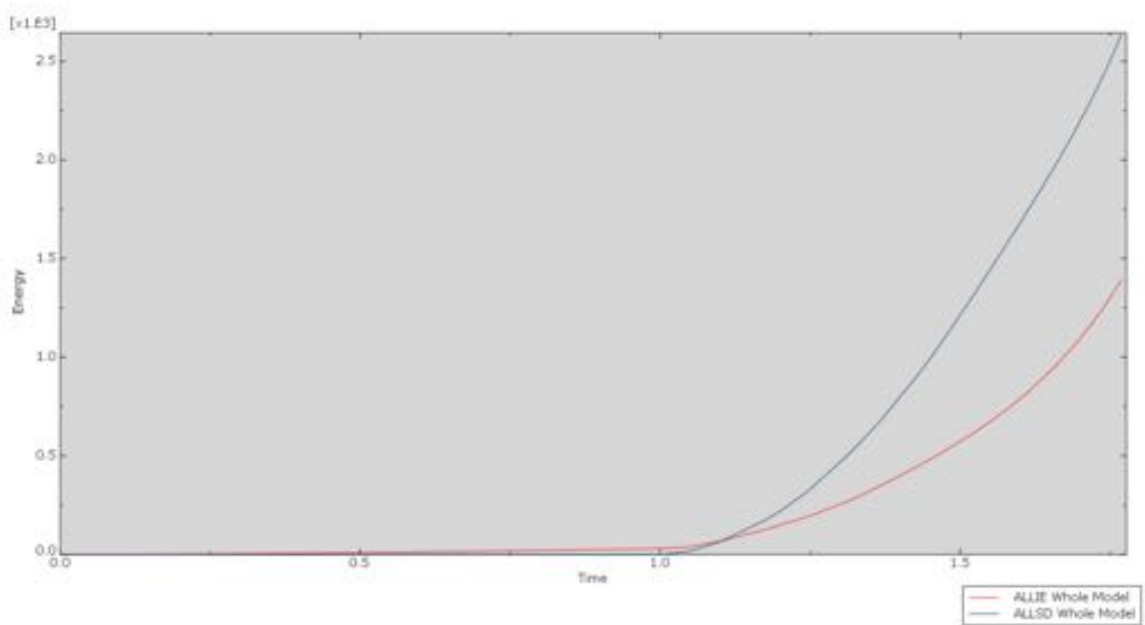


Figure 4-34. Plot Of Total Internal Energy (ALLIE) And Static Dissipation Energy (ALLSD) Vs Time For A Two-Step Simulation (0-1.0 And 1.0-2.0) Where The First Step Uses Automated Damping And The Second Step Uses A User-Defined Constant Damping Coefficient. Simulation Uses The Subject-Specific Gait Exercise Loading Data From Participant 2018Jan19-01.

It is unclear why the simulation is unable to converge when using the automated stabilization scheme. The message file shows distorted elements, a common cause for

convergence failure throughout the course of this study. Further work on understanding how the applied damping factor affects the simulation is recommended. Some preliminary searches have turned up the idea that the rigid bones could be impacting convergence ability. As discussed earlier, modeling bones as rigid surfaces does not impact the pressure experienced by the articular cartilage, but rigid bodies undergoing large rotations and translations can cause convergence problems. Another potential area to study is applying local damping at the contact sites instead of a global damping factor, this should reduce the amount of energy dissipated.

5. CONCLUSION

The goal of comparing control and transtibial amputee tibial cartilage contact pressure developed in three maintenance exercises (gait, cycling and elliptical training) has been partially met. Three participants, one control and two amputees, have physiologically relevant contact pressure results from finite element simulations. These results show that under loads from gait, both tibial compartments experience the largest magnitude contact pressure over the largest area, followed by elliptical and then cycling in descending order of both magnitude of contact pressure and area of application. Between control and amputee participants, the amputee participants had larger magnitude bodyweight normalized contact pressure in both compartments than the control participants in nearly all exercises. The one exception was the medial compartment in the elliptical training load case.

However, because there is not the minimum of three participants in each group, statistical analysis to determine significance of these results is not possible. These results can still be useful in directing future research in this area.

The most impactful limitations of the study are the application of a global damping factor, poor element quality in some bodies, and manual segmentation. The global damping factor would ideally be removed but if it is necessary for convergence, the automatic scheme should be used. This also ties into the second limitation of poor element quality. The global damping factor helps to reduce distortion in the elements, extremely important for this thesis as some of the tetrahedral elements started the simulation distorted. The element quality could be improved through the use of a commercial meshing solution or manual meshing. If manual meshing is used, it is recommended to move to a hexahedral meshing scheme as they are more resistant to excessive distortion than tetrahedral elements. To address inconsistencies in segmentation, automated segmentation based off saturation levels should be explored. While this may not be possible with the current MRIs used in this thesis, the use of a more powerful machine to attain higher quality images would allow for better differentiation of tissues.

The goal of developing a robust, automated schema for physiologically accurate simulations of the TF joint has been advanced but has not yet been finished. The TF is one of the most analyzed joints in the human body due to its impact on quality of life and every new study contributes to a better methodology for the development of an accurate FE model. The results generated from this thesis are another step forward in this decades long project carried out by various researchers and labs. This thesis is novel in its success in outputting multiple converged subject-specific whole-knee finite element models. Along with that, this thesis provides a detailed methodology and all files necessary to replicate the results presented. It is with hope that this schema can be altered and

improved thanks to the exhaustive nature of the described modeling. This author is pleased to present the results contained within and hopes to see further advances in the development of analytical tools, such as subject-specific finite element models, that can be used to help those suffering from debilitating diseases such as early-onset osteoarthritis.

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