Ear pulse waveform parameters during the gradual onset centrifuge training profile

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Executive Summary

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During regular centrifuge training at the Aeromedical Institute the ear photoplethysmogram and the ECG of 15 (candidate) pilots were measured during a gradual onset acceleration profile, without performing an anti-G straining maneuver.
The main goal of the present study was to evaluate the use of the ear pulse waveform (ear pulse) as a potential feedback parameter of a pilot's blood pressure status during accelerations. From the ear pulse waveform, the amplitude of the pulse and the pulse transit time of each pulse were determined.

The results showed clearly that the amplitude of the ear pulse waveform decreased with increasing acceleration level during a centrifuge run. Most trainees showed an initial plateau in the amplitude values with an decreasing amplitude at the end of the exposure. In order to investigate the relationship of the amplitude with each trainee's peripheral light loss (PLL) endpoint, the G_z-levels were converted to percentages of the G-tolerance and fitted with a sigmoidal curve. This curve fitted the individual response accurately (average r=.92). Fitting a response curve across all the data of the trainees gave a R-coefficient of 0.75 and a center of the curve at 80% of the G-tolerance value.
Simultaneously with the decrement of the amplitude, the pulse transit time (PTT) of the ear pulse waveform increased with increasing acceleration level. Most trainees showed an initial slow increase with an accelerated increase at the end of the centrifuge run. Only one subject exhibited a decrease of the PTT values during the exposure to the acceleration profile.

In order to evaluate the use of a parameter as a predictor for the onset of PLL, survival analysis was performed by means of the Kaplan-Meier method for estimating the survival distribution of the length of the 'lead time'. The lead time was defined as the time period starting when the amplitude of the ear pulse waveform was below a certain threshold level up to the moment of PLL.
It was found that an amplitude threshold set at 40% of the amplitude value at 1 G, the average lead time is 9.6 s (with a minimum 2.3 s) and with an amplitude threshold set 10% of the amplitude value at 1 G, the average lead time is reduced to 4.3 s (with a minimum of 0 s) before PLL was reported by the subjects.
If PTT was used as a parameter, a threshold of 120% of the average lead time is 23 s (minimum 11.2 s) and that with a threshold of 180% the average lead time is reduced to 11 s with a minimum of 1.1 s for one subject. Further, it was shown that the experimental results of this study are in agreement with the theoretical models of pulse wave transmission through human blood vessels.

[redacted]

On basis of the earlier extensive work of Wood and coworkers and this study, it is concluded that the ear pulse can be implemented as a feedback parameter for PLL onset. In order to implement the ear pulse waveform as a standard feedback signal during centrifuge training, the reliability of the signal must be evaluated under different G-onset rates and with pilots performing an anti-G straining maneuver. Specially, the delay between the changes in the ear pulse waveform and changes in head level blood pressure must be investigated under higher G-onset rates. So, the ear pulse wave form changes will have to be correlated with continuously measured blood pressure.

Contents

1 Introduction

Pilots in high performance aircraft are frequently exposed to high $+G_z$-forces, causing a redistribution of blood towards the lower extremities. As a result, blood pressure at head level decreases (Agard, 1995; Burns, 1992; Burton, 1991). If the pilot is subjected to high and/or sustained G_z-forces, loss of peripheral vision and/or gray-out will occur as one of the first symptoms. This is called peripheral light loss (PLL) (Agard, 1990). Higher and/or longer exposures may result in a totally reduced visual field (black out) and finally the pilot can experience a G-induced loss of consciousness (G-LOC) (McCloskey et al., 1992; Wood, 1987). This G-LOC incapacitates the pilot for some 10 to 15 seconds, and after recuperation remains disoriented for a certain period (Whinnery, 1988). The tolerance to $+G_z$ can be extended by performing an anti-G straining maneuver, use of an anti-G suit and positive pressure breathing under G (Burns, 1988). However, if the onset rate of the acceleration is too fast or the anti-G straining maneuver is performed inadequately, head level blood pressure may drop critically, resulting in a G-LOC. Until now, no operational system is available to monitor the pilot's head level blood pressure during high acceleration in-flight maneuvering.

Furthermore, (candidate) military pilots are trained in a human centrifuge to optimize the anti-G straining maneuver. During this centrifuge training the trainee is monitored by means of ECG signal and a video system. No objective information is available to the trainee or supervisor about the actual blood pressure or about the effectivity of the trainee's anti-G straining maneuver performance. During the onset of the G_z-acceleration less muscle tension is necessary than during high G_z-forces (Gillingham, 1987). The trainee is given verbal feedback by the supervisor to improve his anti-G straining maneuver and on the basis of his own perception of visual symptoms (e.g., peripheral light loss, grey-out). The instructor gives feedback only based on the trainee's timing of his in- and exhalation, and not on an objective parameter for blood pressure at head level. An objective measuring technique, with a fast response and a low rate of false alarms, is essential for training.

Until now, a limited number of prototype systems were developed to monitor head level blood pressure in a centrifuge, but with limited practical results (Cammarota, 1991). These biofeedback systems were based on the oxidative status of the brain (Glaister, 1988), velocity of blood flow (Chiu et al., 1991), blood pressure at head level (LaCourse et al., 1991; Glaister, 1988) and the arterial ear pulsations (Hebrien 1988; Holewijn e.a. 1994; Jaron et al. 1987; Wood, 1987; Wood, 1990). However, none of these systems were generally accepted as a feedback system in a centrifuge due to several (practical) factors. To be applicable in more than experimental studies the system must be reliable, non-invasive, non-interfering and small (so that it can be worn under a pilot helmet) and it must sustain high G_z loads that occur during flight (Whinnery, 1989).

At the Netherlands Aeromedical Institute several experiments were dedicated to develop a demonstrator for a head level blood pressure feedback system. This system is based on the arterial ear pulsations, a so-called photo-electric plethysmographic ear pulse waveform monitor.

The results of previous photo plethysmography studies showed that the amplitude of the photo-plethysmogram of the ear (Holewijn et al., 1994; Wood et al. 1989, 1990; Wood and Sturm, 1989; Wood et al. 1990) and of the temporal artery on the head (Hrebien, 1988; Jaron et al. 1987; Wood 1987,1990; Wood and Sturm, 1989) decreased with increasing G_z-strain. Furthermore, these studies showed that the amplitude of the ear pulse waveform decreased to zero (no pulsations) a few seconds before the moment of peripheral light loss (PLL) onset.

The main goal of the present study was to evaluate the use of the ear pulse waveform as a potential feedback parameter of a pilot's head level blood pressure during accelerations. From the ear pulse waveform, the amplitude of the pulse and the pulse transit time of each pulse were determined. Data were collected during regular centrifuge training at our institute, in order to evaluate its reliability. The hypothesis was that the amplitude and the pulse transit time (PTT) of the ear plethysmogram could be used as predictors for the moment of PLL. It must be mentioned that the reliability and accuracy of this subjective determination of G_z-tolerance can be low due to factors such as fatigue, maliciousness and disinterest (Coburn, 1970). However, due to the manner in which regular training sessions are registered, no other parameters (e.g., non-invasive blood pressure measurement) could be used as a reference. This project was carried out under contract of the Dutch Ministry of Defense.

2 Methods and materials

2.1 Subjects

During centrifuge training 15 F-16 NATO pilots were measured. These pilots followed a regular training to the Aeromedical Institute at Soesterberg in the Netherlands. The pilots were medically qualified as fit for flying and had a mean age of 25 years (range 23-26). No other subject characteristics are collected during the regular training. The data of two trainees were excluded from further analysis due to low data quality.

2.2 Physiological measurements

During the profiles the ear pulse waveform and the electrocardiogram (ECG) were recorded. After hardware filtering the signals were A/D converted and stored on disk with a sample frequency of 125 Hz per channel on a PC by means of the Codas Data Acquisition System (Dataq Instruments Inc., Ohio, USA). The data were further processed offline. The physiological and the G_z signals were smoothed with a 3 points moving average filter.

ear pulse waveform

Blood has a higher absorption coefficient for (near) infra red light than tissue. Therefore, if blood volume in the ear pinna increases, less (near) infra-red light will be transmitted through the ear pinna, and the amplitude of the photo-plethysmographic signal will drop. It must be noted that photo-electric plethysmography cannot directly measure volumetric changes, due to the measuring method.

In his literature review, Challoner (1979) concluded that the pulsatile component is correlated with blood volume variations and that the D.C. component is related to the amount of total (pooled) blood volume and tissue between the sensor and the light source (Fig. 1).

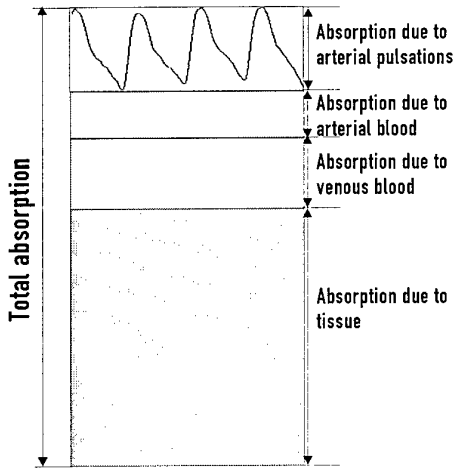

Fig.1. Schematic representation of the absorption components of a photo-plethysmographic signal. In accordance to the literature, the photo-plethysmographic signal is inversely presented.

The amplitude of the transmission photoplethysmographic signal contains a small alternating (pulsatile) opacity component, and a large steady state (D.C.) component (Barron et al. 1993; Kamal et al. 1989). This pulsatile component in the ear opacity waveform is composed of a heart rate dependent photoplethysmogram opacity pulsation and a slower, ventilation dependent, amplitude variation (Dorlas and Nijboer, 1985). The ventilation dependent amplitude variations are caused by intrathoracic pressure changes during respiration. These pulsatile pressure changes are mainly transmitted to the periphery in the venous pressure compartment of the cardiovascular system. The ventilation dependent plethysmographic amplitude variations are slower than the heart beat correlated pulsations in the photo-plethysmographic signal. In order to minimize temperature induced vasoconstriction (Dorlas and Nijboer, 1985), care was taken to measure at an ear skin temperature of 33-34 °C, obtained due to wearing of ear cuffs.

These blood volume pulsations (ΔV) in the peripheral arterial bed depend on the changes of the pulse pressure (ΔP) and the vascular distensibility (D), according to a relationship described by Burton(1972) as:

$$\Delta V = D * \Delta P \qquad (1)$$

In the peripheral vascular bed, the distensibility depends mainly on the tone of the vascular smooth muscles, controlled by the autonomic nervous system. In several peripheral tissue regions, the effect of the autonomic system on the amplitude of the plethysmogram may completely dominate over the opposite effect of pulse pressure changes (Dorlas and Nijboer 1985).

Changes in the autonomic regulation due to temperature or psychological stress can influence significantly the amplitude of the plethysmogram in absence of changes in pulse pressure, and are clearly visible in the finger arterial bed(Challoner, 1979; Dorlas and Nijboer, 1985; Hertzman and Dillon, 1939; Nijboer et al. 1981; Stern, 1974). In contrast to the finger photoplethysmogram, the amplitude of the plethysmogram of the ear is less sensitive to psychological and physiological stimuli (Dorlas and Nijboer, 1985; Hertzman and Dillon, 1939; Nijboer et al., 1981; Stern, 1974), but mostly dependent on perfusion pressure (Midttun and Sejrsen, 1996).

The continuous ear pulse waveform was measured with a custom-built transilluminal photoplethysmograph, consisting of a Telefunken Silicon Photo PIN diode (BPW34) and a Silicon Planar PN Photovoltaic Cell (BPW35). A Fleishman 9808 lightbulb was used as light source and filtered with a high-pass filter of 800 nm (near infrared). At this wavelength the tissue penetration depth of the light source is around 1.5 mm, and the pulsations of the larger vessels will be measured (Ugnell and Öberg 1995).
The signal was bandpass filtered (0.1-25 Hz) with an analog 6dB/oct hardware filter before A/D conversion and digital storage on disk. The ear pulse waveform was off-line processed with data processing software package (Windaq 1.68, Dataq Instruments Inc). The ear pulse waveform was digitally bandpass filtered between 0.6 and 12 Hz by means of Fourier inverse filtering.

Of the ear photoplethysmogram the amplitude, or also called the ear opacity pulse, was used for further analysis.
Besides the ear pulse amplitude changes also the pulse transit time of each pulse was determined (Fig. 2). The onset of each ear pulse waveform was determined by means of peak detection on the second derivative of the ear pulse waveform signal (Chiu et al. 1991). This foot of the plethysmogram waveform is almost free of pressure wave reflection and can be determined reliably (Mitchell et al. 1997). This point of the waveform was defined as the valley of an ear pulse waveform. The peak value of the ear pulse waveform was defined as the maximal value of the pulse. The amplitude of the ear pulse waveform was calculated as the difference between the peak and the valley (Fig. 1).

As the ear pulse waveform is not a calibrated signal, all ear pulse amplitude data are presented as relative values of the reference value at rest. The mean value of the amplitudes of the ear pulse waveform of each subject, recorded during the first 30 s of the centrifuge training profile, was used as a reference value.

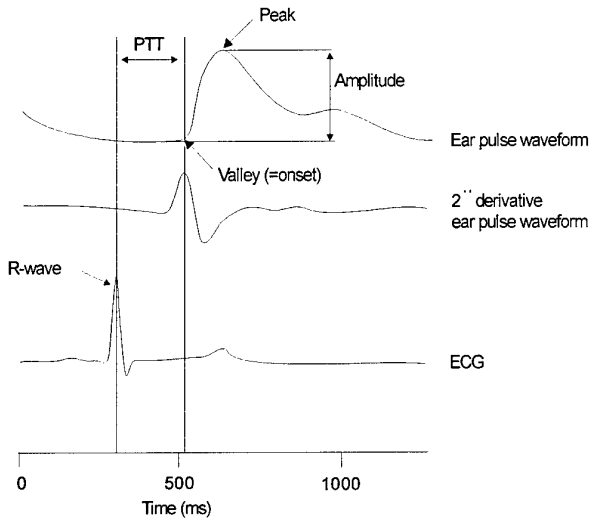

Fig. 2. The definition of the amplitude and the pulse transit time (PTT) of the ear pulse waveform.

The pulse transit time (PTT, ms) of the ear pulse waveform was defined as the time between the R-wave of the ECG and the onset of the ear pulse waveform. A correction of minus 40 ms was applied to subtract the time period of the isovolumetric period (Eliakim et al. 1971; Weltman 1964). The PTT calculated from the ear plethysmogram showed the highest repeatability as compared to the finger or toe plethysmogram (Jago and Murray 1988).

Similar to the transformation of the amplitude, all the calculated PTT values were transformed to relative units as compared with the mean PTT-value of the first 30 s before the onset of the G_z-training profile.

ECG

The R-waves were derived from the ECG signal using a Mennen Cardiograph (Model 700). After skin preparation three electrodes were placed on the right and left clavicular line, directly below the clavicle and on the 6th intercostal space on the left midclavicular line. The ECG signal was bandpass filtered between 0.1-100 Hz before A/D conversion.

2.3 Experimental procedure

During the regular training of (candidate) F-16 pilots, measurements were done on the pilots who volunteered to participate, after being informed about the purpose of the study. During the training at the Aeromedical Institute the relaxed-G profile is always the first profile which a trainee is subjected to, after a classroom instruction.

The used 'relaxed-G profile' had an onset rate of 0.1 $G \cdot s^{-1}$ and was stopped by pilot at the moment of experiencing gray-out or peripheral light loss (PLL). The G-level at this point was indicated as his relaxed G-tolerance level.

2.4 Statistics

Nonlinear quadratic and linear regression analyses were performed between the different physiological variables and the G_z level with the statistical software package Systat (version 7.01, SPSS inc). A significance level of $P \leq 0.05$ was used. In the figures the data points were fitted with a non-linear line, using the distance weighted least square (DWLS) method.

In order to evaluate the use of a parameter as a predictor for the onset of PLL, survival analysis was done by means of the Kaplan-Meier method for estimating the survival distribution of the length of the 'lead time'. The lead time was defined as the time period starting when the amplitude of the ear pulse waveform was below a certain threshold level up to the moment of PLL (Fig. 3).

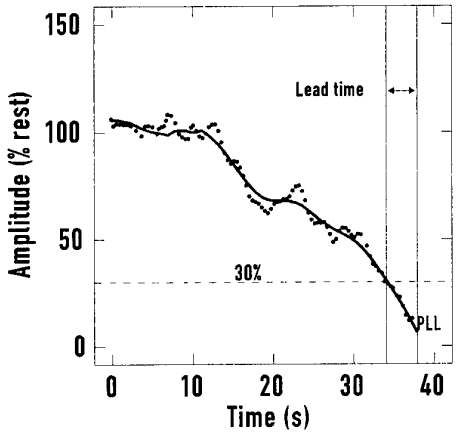

Fig. 3. The calculation of the lead time for an amplitude level of the ear pulse waveform set at 30%. The non-linear line is fitted with the DWLS method.

As can be seen in Fig. 3 the higher the level is set, the longer the lead time will be, until the moment of PLL occurence. However, the longer this period, the higher the uncertainty in the prediction will be. Lead times were calculated for the amplitude levels starting at 40% down to 5% in steps of 5%. Analogue lead times were calculated for PTT levels of 120% up to 180%, in steps of 10%. The estimated survival distributions were statistically compared for significant differences, using the Mantel-Haenszel log-rank test (Systat 7.0, 1997).

3 Results

3.1 General results

The average G_z-tolerance during the relaxed G-profile was 4.9 G, but considerable variance was found (Fig. 4). Two subjects had a G-tolerance lower than 3.5G.

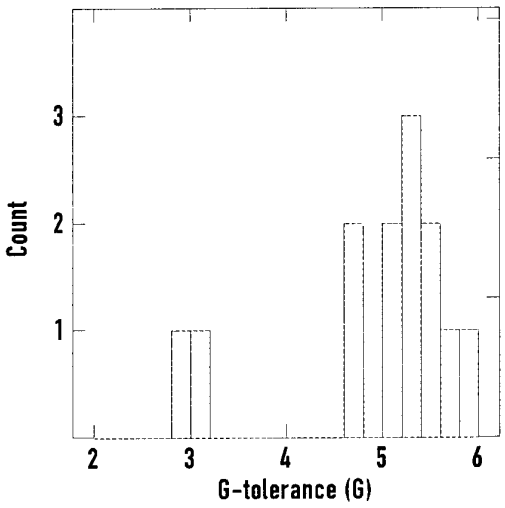

Fig. 4. The distribution of the G-tolerance during the 'relaxed G' profile.

No training run was prematurely stopped due to arrhythmia or motion sickness or other reasons. The mean heart rate increased from 120 beats·min^{-1} (SD: 23) at the start of the centrifuge training profile up to 147 beast·min^{-1} (SD: 17) at the moment of profile abortion due to occurrence of peripheral light loss (PLL).

The interbeat intervals (IBI) calculated from the R-waves of the ECG, were highly correlated with the IBI's calculated from the peak and valley of the ear pulse waveform (Fig. 5), the latter having the highest correlation with the IBI's derived from the ECG.

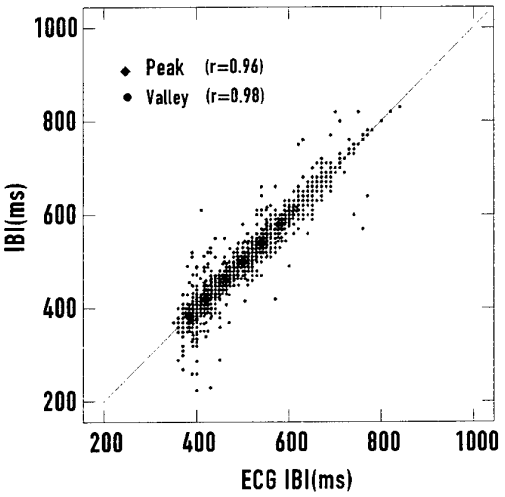

Fig. 5. The relationship between the interbeat interval (IBI,ms) calculated form the ECG and the IBI calculated form the earpulse waveform peaks and valleys

Along with the changes in heart rate, a general increment in the pulse transit time (PTT) was found in combination with a decrement of the amplitude of the ear pulse waveform (Fig. 6).

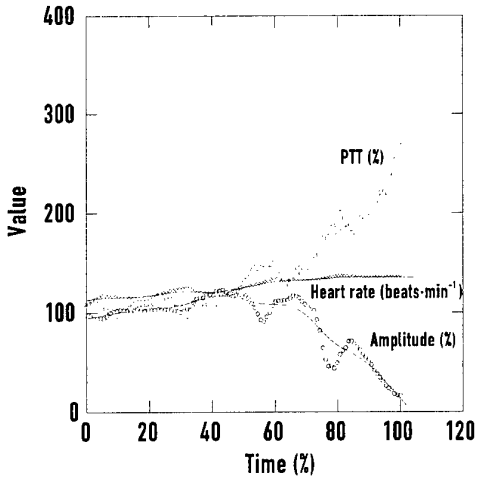

Fig. 6. A typical example of the changes in the heart rate (beats·min^{-1}), PTT (% rest value) and the amplitude (% rest value) of the earpulse waveform in time (% tolerance time).

A sinusoidal variation can be seen on the response of the amplitude and PTT which presumably correlates with the respiration.

3.2 Amplitude of the ear pulse waveform

The amplitude of the ear pulse decreased with increasing acceleration level (Fig. 7). Most trainees showed an initial plateau in the amplitude value and a decreasing amplitude at the end of the exposure.

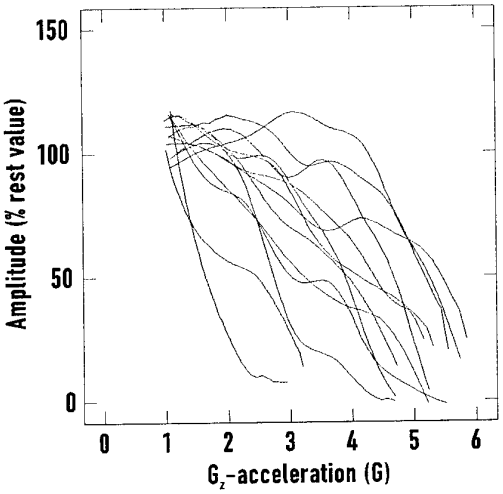

Fig. 7. The decrement of the amplitude of the earpulse against the G_z-level. The individual curves are fitted lines (DWLS method) and end where the trainee reached his G-tolerance.

In order to compare the relationship of the amplitude with each trainee's PLL endpoint (the relaxed G-tolerance point), the G_z-levels were converted to percentages of the G-tolerance. Each trainee's curve was also fitted to an inverted S-shaped relation (1) ending at the point of 100% G_z-stress (Fig. 8, left panel). Fitting a curve over all the data resulted in a mean response of the amplitude of the ear pulse in response to increasing G_z stress (Fig. 8, right panel).

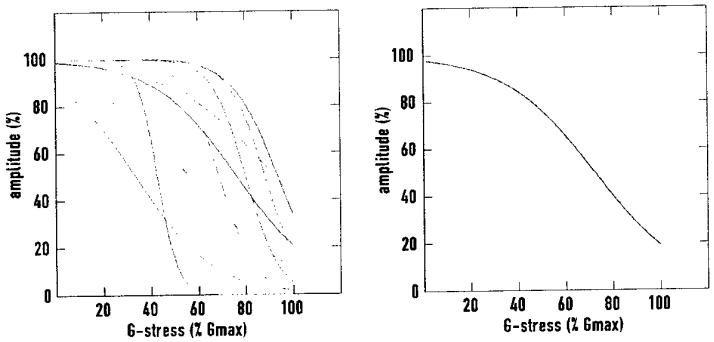

Fig. 8. The fitted relationship between the amplitude (% rest value) of the ear pulse waveform and the G_z-stress level (% of G-tolerance) for all the trainees (left panel) separately, and fitted to the data of all the trainees (right panel).

The amplitude decay the with increasing G_z -stress was fitted with the sigmoidal curve:

$$amplitude(\%) = 100 * \frac{1}{1 + e^{(slope * (G_{stress} - center))}} \qquad (2)$$

The 'slope' variable represents the steepness of the curve and the variable 'center' gives the center point (of the 50% value) of the sigmoidal curve. Fitting this nonlinear curve to each trainee's response yielded the data presented in Table 1.

Table 1. Nonlinear fit coefficients of the relationship between the amplitude of the ear pulse waveform and the G-stress (% Gmax) and the correlation coefficient for 13 subjects.

	Mean (n=13)	SD
slope	0.09	0.06
center	69	19
correlation coefficient	0.92	0.05

This sigmoidal curve could fit the trainee's individual response accurately as can be seen from the correlation coefficients.
Fitting an average response curve across the data of all trainees resulted in a correlation coefficient of 0.75 and a 50% center point of the curve at 78% of the G-tolerance value.

3.3 PTT of the ear pulse

The PTT(% rest value) of the ear pulse increased with increasing acceleration level (Fig. 9). Most trainees showed an initial slow increase with an accelerated increase at the end of the centrifuge run. One subject exhibited a decrease of the PTT after an initial increase during the exposure to the acceleration profile.
In order to compare the relationship of the amplitude with each trainees PLL endpoint (G-tolerance point), the G_z-levels were converted to percentages of the G-tolerance. Each trainee's curve was fitted with an exponential curve, ending at the point of 100% G_z -stress (Fig. 10).

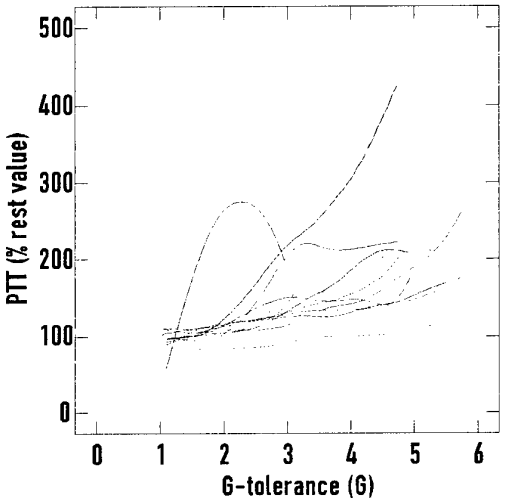

Fig. 9. The relationship between the PTT of the earpulse and the G_z-level. The individual data point are fitted with a line (DWLS method) and end where the trainee reached his G-tolerance.

The individual responses could be fitted reasonably with a quadratic regression line. The average correlation was 0.85 (range 0.22-0.95), but there was a large inter-individual variation in the response (Fig. 10, left panel). Fitting a curve over all the data gave a correlation coefficient of 0.35.

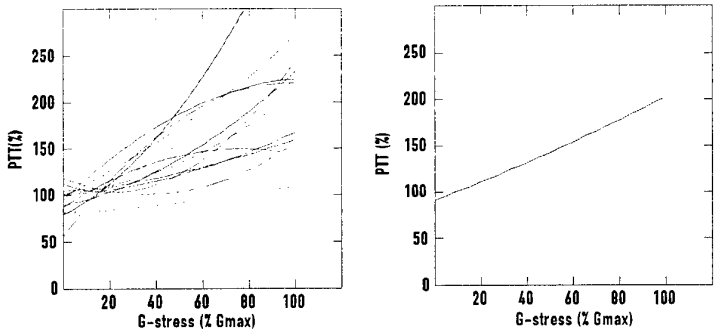

Fig. 10. The fitted relationship between the PTT (% rest value) of the earpulse wave form and the G-stress level (% of G-tolerance) for all the trainees (left panel) and averaged across all trainees (right panel).

3.4 Heart rate

All the subjects exhibited an increase of the heart rate with increasing G-stress, however there was considerable variation in the starting level (Fig. 11). Furthermore, some trainees showed a linear increase and other trainees showed a superimposed, respiration-dependent variation. The average linear correlation coefficient with G_z-stress (% G_{max}) was 0.75 (range:0.36-0.97).

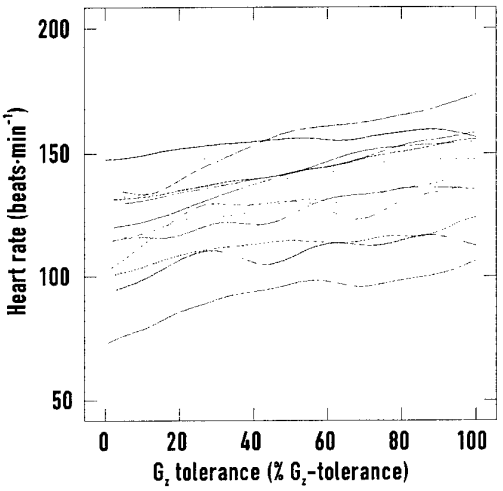

Fig. 11. The fitted relationship (DWLS method) between the heart rate (beats · min^{-1}) and the G-stress level (% of G-tolerance) for all the trainees.

There was a statistically significant interaction between the subject and the starting heart rate value and between the subject and the increase in the heart rate (slope of the curve). This is visible in Fig. 11 as large differences in the initial heart rate levels and differences in the amount of increase in heart rate per unit increase of relative G_z-level.

3.5 Relationship between amplitude, PTT and heart rate

The amplitude of the ear pulse waveform changed inversely with the PTT value. With increasing G_z -levels, amplitude values decreased from 100% to around 0% in combination with increasing PTT values (Fig. 12). An asymptotic inverse function (amplitude=a+b·PTT^{-1}) correlated reasonably well with the data, with an average correlation coefficient of 0.89 (range 0.68-0.96).

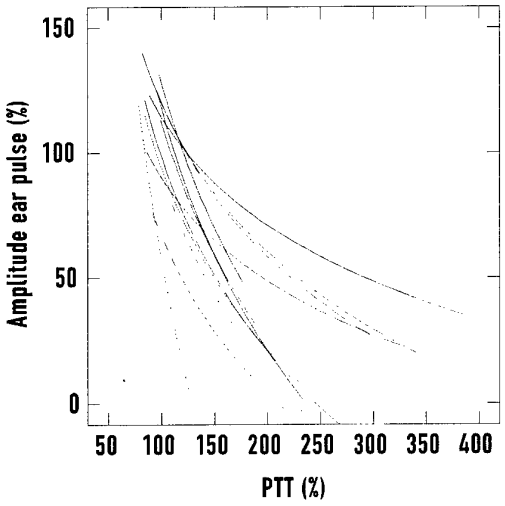

Fig. 12. The fitted relationship between the amplitude (% rest value) and the PTT (% rest value) for all the individual trainees.

The changes in heart rate could be linearly correlated with the changes in amplitude and PTT (Fig. 13).

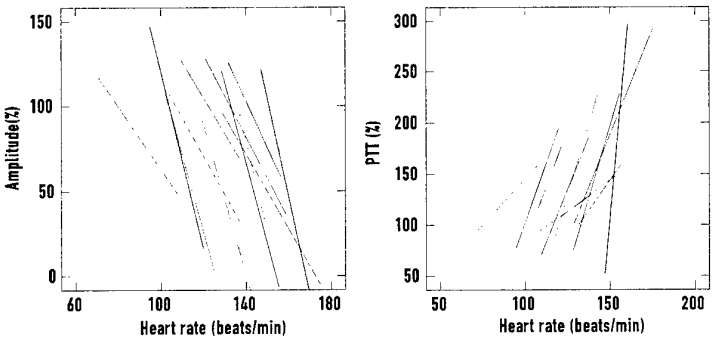

Fig. 13. The fitted linear relationship between the heart rate and the amplitude of the earpulse wave form (left panel) and between the heart rate and the PTT of the earpulse wave form (right panel).

The correlation between the heart rate and the amplitude (mean: 0.84, range: 0.60-0.95) was slightly higher then the correlation between the heart rate and the PTT (mean: 0.74, range: 0.41-0.92).

3.6 Prediction of PLL with lead times

On basis of the calculated lead times of the 4 different amplitude and PTT threshold levels, survival distributions were estimated with the Kaplan-Meier method (Fig. 14). In this analysis lead time is used as a survival time.

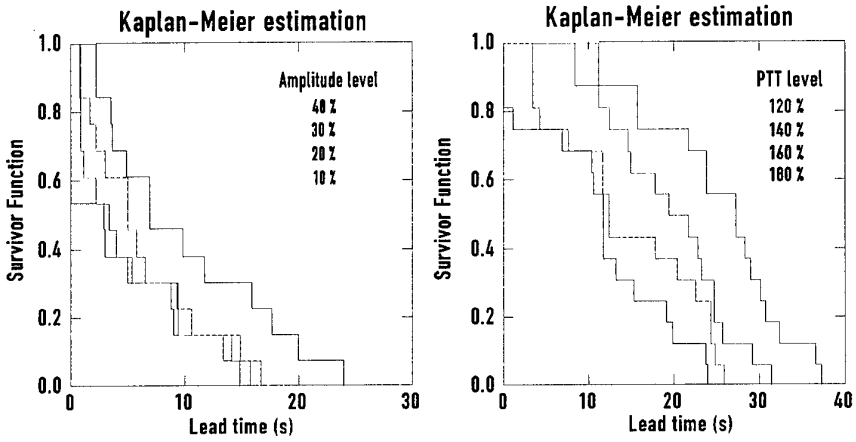

Fig. 14. The estimated Kaplan-Meier survival distributions of the different amplitude (left panel) and PTT (right panel).

It can be seen that the higher the amplitude threshold levels are chosen, the longer the lead times are. In order to compare the 4 different threshold levels, the mean and the minimum lead times were derived from the estimated distribution curves (Table 2). It can be seen that with an amplitude threshold set at 40% the average lead time is 9.6 s (with a minimum 2.3 s) and with an amplitude threshold set at 10% the average lead time is reduced to 4.3 s (with a minimum of 0 s) before PLL was reported by the subjects. The Mantel-Haenszel log-rank test revealed that the survival distribution was only statistical significant different between the 10% and the 40% amplitude threshold level.

Table 2. Estimated mean and minimum survival times (s) with 4 different amplitude threshold levels.

Amplitude threshold level (% of the initial value at 1 G)	Survival time (s)	
	mean	minimum
10	4.3	0
20	5.0	0.8
30	6.3	0.8
40	9.6	2.3

In order to compare the effect of different PTT threshold levels, the average and the minimum lead times (Table 3) were derived from the estimated distribution curves (Fig. 14, right panel). In Table 3 it can be seen that with a PTT threshold of 120% the average lead time is 23 s (minimum 11.2 s) and that with a threshold of 180% the average lead time is reduced to 11 s (minimum 1.1 s). The Mantel-Haenszel log-rank test revealed that the survival distribution of the 120 % threshold was statistically significant different from the other 3 threshold levels.

Table 3. Estimated survival times (mean and minimum) with 4 different PTT threshold levels.

PTT threshold level (% of the initial value at 1 G)	Survival time (s)	
	mean	minimum
120	23	11.2
140	18	11.2
160	13	3.4
180	11	1.1

4 Discussion

The main goal of the present study was to evaluate the use of the ear pulse waveform (ear pulse) as a potential feedback parameter of a pilot's head level blood pressure during accelerations.
The general results indicate that the pulse amplitude (%) of the ear pulse waveform decreased with increasing acceleration levels during a centrifuge run. Most trainees showed, after an initial plateau, a decrease in the amplitude values with increasing G_z-levels and at the end of the exposure a leveling off of this decrement. The individual data points could be adequately fitted with a sigmoidal response curve (mean r=.92). Fitting a sigmoidal response curve across the data of all trainees gave a correlation coefficient of 0.75.
Simultaneously with the decrement of the amplitude, the PTT (%) of the ear pulse increased with increasing acceleration levels. Most trainees showed an initial slow increase with an accelerated increase at the end of the investigated centrifuge run. Only one subject exhibited a decrease of the PTT after a sharp increase during the exposure to the acceleration profile. The individual responses could be fitted reasonably with a quadratic regression line. The average correlation was 0.85 (range 0.22-0.95), but there was a large inter-individual variation in the response. Fitting a quadratic response curve over all the data gave a correlation coefficient of 0.35.
The changes in heart rate were found to be linearly correlated with the changes in the relative amplitude and PTT. The correlation between the heart rate and the amplitude (mean: 0.84, range: 0.60-0.95) was slightly higher than the correlation between the heart rate and the PTT (mean: 0.74, range: 0.41-0.92).

In order to evaluate the use of a parameter as a predictor for the onset of PLL, survival analysis was performed by means of the Kaplan-Meier method for estimating the survival distribution of the length of the 'lead time'. The lead time was defined as the time period starting when the amplitude of the ear pulse waveform was below or the PTT value was above a threshold level up to the moment of occurrence of PLL.
It was found that, with an amplitude threshold set at 40% of the amplitude value at 1 G, the average lead time is 9.6 s (with a minimum 2.3 s) and with an amplitude threshold set at 10% of the amplitude value at 1 G, the average lead time is reduced to 4.3 s (with a minimum of 0 s) before PLL was reported by the subjects
If using different thresholds for PTT lead time calculations, a threshold of 120% resulted in an average lead time is 23 s (minimum 11.2 s) and with a threshold of 180% resulted in an average lead time of 11 s (minimum of 1.1 s).

The validity of using the photoplethysmogram of the ear as an indicator of the head level blood pressure, is of major importance. The ear lobe tissue is supplied by blood by the internal and external carotid arteries. Photoplethysmography is used in several fields of cardiovascular research. The main fields are non-invasive assessment of left ventricular performance (Bernardi et al. 1984; Chirife and Spodick, 1972; Haffty et al. 1983; Kobayashi et al. 1978; Nakamura et al. 1980; Quarry-Pigott et al. 1973; Sugiura et al. 1980), measurement of vasomotor function (Barron et al. 1993), peripheral resistance (Haffty et al. 1983), diagnosis of venous diseases (Fronek 1995, Sherebrin and Sherebrin, 1990), and non-

invasive quantification of arterial elasticity (Callanghan et al. 1986; Geddes et al. 1981; Gribbin et al, 1976; Jago and Murray, 1988; Pruett et al., 1988).
A basic question concerning the use of transmission ear photoplethysmography as used in the present study is how the blood volume changes (pulsations) in the vascular bed in tissue are reflected in the ear pulse waveform signal. Blood has a higher absorption coefficient for (near) infra-red light than tissue. Therefore, if blood volume in the ear pinna increases, less (near) infra-red light will be transmitted through the ear pinna, and the amplitude of the photo-plethysmographic signal will drop.
According to Jespersen and Pedersen(1986) the relationship between the output of a transmission photoplethysmograph and the amount of red cell density can be adequately described with a linear function(r^2=0.86), although a logarithmic gives an even higher correlation(r^2=0.99). The results of several other studies on blood volume changes by means of the mercury-in-rubber strain-gauge-finger plethysmography technique clearly indicated that photo-electric transmission plethysmography gives identical results (Kramer et al. 1963; Nijboer et al. 1983, Samuelsson et al. 1996; Zijlstra and Mook 1962). Furthermore, the results of several studies indicate that the ear photoplethysmogram has a high correlation with the carotid blood pressure pulse (Bernardi et al. 1984; Lance and Spodick 1977;Quarry-Pigott et al. 1973). Thus, it is concluded that the amplitude of transmission photo-plethysmography is correlated with the blood volume changes in tissue due to blood pressure effects.

The general effects of G_z-acceleration on the ear opacity pulse and on the PTT found in this study are in agreement ith the results of studies on the effects of G_z-acceleration on blood pressure and with the results of the studies on the effects of blood pressure on blood vessel properties.
It is well known that the cross-sectional area of a blood vessel is non-linearly related to the blood pressure (Giezeman 1992; Gizdulich and Wesseling 1988; Wesseling et al. 1993;Wesseling et al. 1995). From the arctangent relationship between the cross-sectional area (CSA) and blood pressure as reported by Wesseling et al. (1993), the global trend of the effect of blood pressure on the CSA of a vessel can be estimated (Fig. 15). It can be seen that the CSA of a vessel increases in a S- shaped manner (Fig. 15, left panel). The steepness of the curve and the absolute values are dependent on the type of vessel and on the contraction status of the smooth muscle in the wall (Giezeman 1992;Wesseling et al. 1995). These volume changes of a vessel, due to a blood pulse pressure wave in the vessel, are dependent on the actual stretch of the vessel. This is reflected in the compliance parameter of a blood vessel, representing the elastic property of a vessel (Fig. 15, left panel). Assuming a cylindrical blood vessel the changes in diameter can be simply derived from the changes in CSA . For simplicity the slight hysteresis in the relationship between vessel diameter and blood pressure as well as the effects of smooth muscle in the vessel's wall are neglected (Giezeman, MacWilliams et al. 1998; Wesseling et al. 1995).
Combining the relationship between compliance and blood pressure with the changes in blood pressure and changes in pulse pressure with increasing G_z-acceleration, the pulsatile diameter changes of a vessel can be calculated, assuming that the mean blood pressure decreases with about 22mHg per G (Burns 1992; Burton 1991; Burton et al. 1974; Howard 1981) and that the pulse pressure decreases with about 10 mmHg per G_z increase (Holewijn et al. 1999).

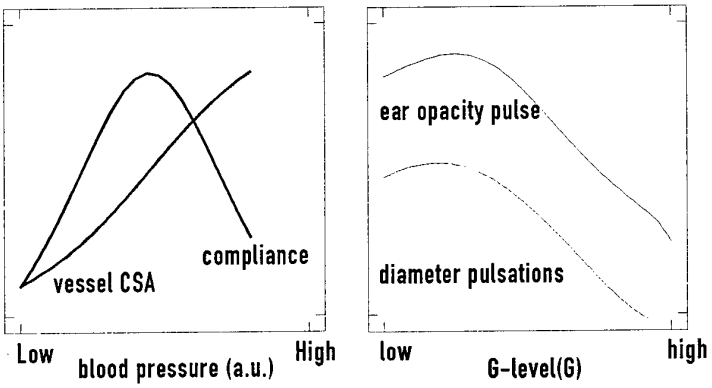

Fig. 15. The estimated relationship between the diameter and the compliance of a vessel and blood pressure (model data Wesseling et al. 1993) (left panel). The estimated relationship between the blood vessel diameter pulsations and ear opacity pulse (=transmitted light) and G_z-level (right panel).

Furthermore, the changes in the pulsatile component of the ear transmission photo-plethysmogram can be derived from the pulsatile changes in the diameter of a blood vessel, using the relationship between the light transmission and the thickness of the blood layer, as described by the Lambert-Beer law (Cenjar et al., 1992; Cui and Otrander 1990). This law (3) describes the absorption of monochromatic light in a non scattering solution, in which I_o and I are the incident and transmitted light intensities respectively, through a cuvette layer of depth D containing a homogeneous solution of concentration C, with an absorption coefficient ε.

$$I = I_0 10^{-\varepsilon CD} \qquad (3)$$

Other models (Twersky's or Kubelka-Munk's model) incorporate the effect of light scatter arising from the refractive index discontinuity at the plasma/erythrocytes interface on the transmitted light intensity (Fine and Weinreb, 1995; Kock and Tarassenko, 1993), but for quantification of the effects of G_z-accelerations on blood pressure, the Lambert-Beer model is assumed to be accurate enough, as transmission photo-plethysmography is less sensitive to the effect of light scattering than reflectance photo-plethysmography (Graaf et al. 1995). Fitting to formula (3) the data of Cejnar et al. (1993) concerning the extinction coefficient of infra red light in tissue, the changes in the transmitted light intensity due to pulsatile diameter changes of a blood vessel are estimated (Fig. 15, right panel). Although the absolute values of the relationship between the ear pulse amplitude and the G_z-level depend on the values of theparameters used, it can be seen that the changes in the relationship resemble the data in Fig. 7. Thus, it is concluded that the experimental results of this study are in agreement with the theoretical models of light transmission through thin walled human blood vessels.

Analogous to the theoretical basis for the changes in the ear pulse amplitude during G_z-accelerations, the changes in PTT of the ear pulse waveform signal will be derived. It is well-known that changes in blood pressure affect the pulse wave velocity (PWV). It should be stressed that PWV is not the same as blood velocity, which is the velocity at which a certain blood volume is displaced in a vessel. PWV values are always higher than blood velocity values (Milnor, 1989; Tardy et al., 1991).
PWV is calculated from the pulse transit time (PTT) of a blood pressure wave and the distance (l) between two root-sharing measuring points in the cardiovascular system according to (4):

$$PWV = \frac{l}{PTT} \qquad (4)$$

Depending on the investigated artery or species, a linear or exponential relationship have been found between the PWV and the mean arterial blood pressure (Callaghan et al, 1986; Pruett et al. 1988; Chiu et al. 1991; Gedes et.al 1981; Gribbin et al, 1976; Jago and Murray 1988, Lu et al. 1992; Steptoe et al.,1976; Tardy et al. 1991) (Fig.16).

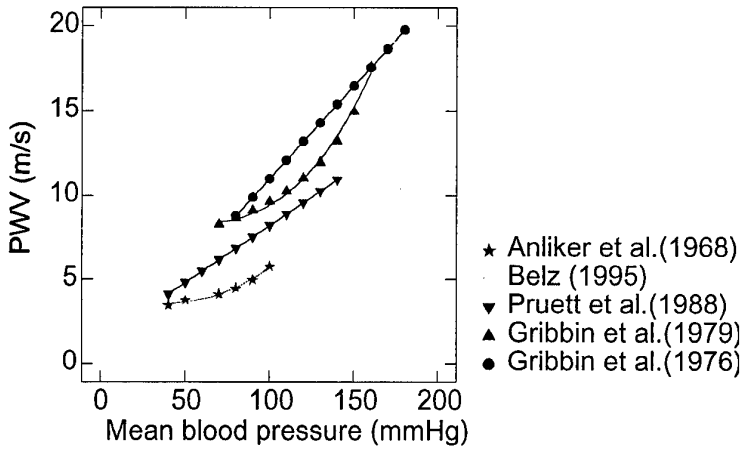

Fig. 16. The relationship between the PWV (m/s) and the mean arterial blood pressure (mmHg) found in the aorta of dog (Anliker et al. 1968; Pruett et al., 1988), human brachialis artery (Gribbin et al., 1976,1979) and human aorta (Belz, 1995).

Furthermore, it has been shown that PWV is mainly related to blood pressure and not to pulse frequency (Callaghan et al. 1984). Schimmler (1965) showed also that the PWV is also affected by the age of the person, as a consequence of changed vessel's elasticity. The PWV calculated from the ear plethysmogram showed the highest repeatability as compared to the finger or toe plethysmogram (Jago and Murray 1988).

The effect of pressure on pulse pressure wave propagation in thin-walled elastic tubes was already described in 1878 by Moens and Korteweg. The relation between the PWV and Young's modulus of elasticity (E) of an arterial wall can be described by the Moens and Korteweg equation(5) as:

$$PWV = \sqrt{\frac{Eh}{2r\rho}} \tag{5}$$

with
E = Young's elasticity modulus (N/m^2)
h = vessel wall thickness (m)
ρ = blood density (kg/m^3)
r =radius of the blood vessel's lumen(m)

When blood pressure increases, the wall thickness (h) decreases and the diameter of the blood vessel increases. This should result in a decrease of the PWV, but it is overshadowed by the opposite effect of the blood vessel's elasticity changes due to blood pressure changes. The elastic modulus (E) of an artery is not a constant but increases exponentially with increasing blood pressure, according to (6):

$$E = E_0^{\alpha BP} \tag{6}$$

with
E_0 = zero pressure modulus (N/m^2)
α =coefficient dependent on vessel type (arteries ± 0.017)
BP=blood pressure (mmHg)

Therefore, PWV will decrease with decreasing blood pressure due to the increase in E, overshadowing the effects of a decrease in vessel thickness and increase in a vessel diameter on PWV (Pruett et al., 1988).

$$PTT = \frac{l}{\sqrt{\frac{E_0^{\alpha * BP} h}{2r\rho}}} \tag{7}$$

Combining equations(4), (5)and (6) yields formula (7) indicating an inverse exponential function between the PTT and the BP. As the specific mass of blood is close to 1 ($\rho \approx 1.03$ kg·m^{-3}) formula (7) can be simplified to (8):

$$PTT = \frac{l}{\sqrt{\frac{E_0^{\alpha * BP} h}{2r}}} \tag{8}$$

Combining formula (8) with a mean blood pressure decrease of 22mHg per G (Burns 1992; Burton 1991; Burton et al. 1974; Howard 1981) and with mechanical vessel characteristics according to Giezeman (1992) a general relationship between the PTT and G_z-level can be derived (Fig. 17).

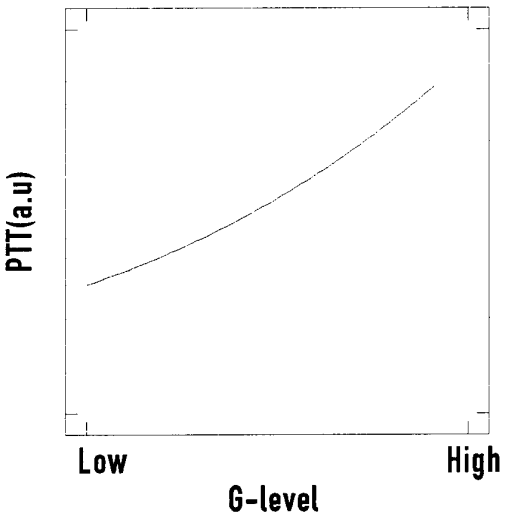

Fig .17. The theoretical relationship between the G-level and PTT based on formula (8).

Although in absolute terms the values of the relationship between PTT and G_z-level depends on the parameters used, it can be seen that the curve shown resembles reasonably well the measured data in Fig. 10. Thus, it is concluded that the experimental results of this study are in agreement with the theoretical models of pulse wave transmission through human blood vessel.

5 Conclusions

On basis of the results of the earlier extensive work of Wood and coworkers and of this study it is concluded that the ear pulse can be implemented as a feedback parameter for PLL onset. In order to substantiate the threshold levels investigated in this study a larger population will have to be recorded to accommodate the total range of variation. Furthermore, the relation of the changes in the ear pulse waveform with head level blood pressure changes should be further investigated.

In order to implement the ear pulse waveform as a standard feedback signal during centrifuge training further research will have to evaluate the reliability of the signal under different G-onset rates and with pilots performing an anti-G straining maneuver. Specially, the delay between the changes in the ear pulse waveform and changes in head level blood pressure under higher G-onset rates requires further investigation.

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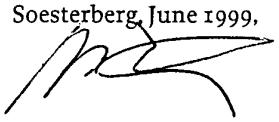

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Ear pulse waveform parameters during the gradual onset centrifuge training profile

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15 ABSTRACT (MAXIMUM 200 WORDS, 1044 BYTE)

During regular centrifuge training at the Aeromedical Institute the ear photoplethysmogram and the ECG of 15 (candidate) pilots were measured. The main goal of present study was to evaluate the use of the ear pulse waveform (ear pulse) as a potential feedback parameter of a pilot's head level blood pressure during accelerations. From the ear pulse waveform, the amplitude of the pulse and the pulse transit time (PTT) of each pulse were determined. The results showed clearly that the amplitude of the ear pulse waveform decreased and the PTT increased with increasing acceleration level during a centrifuge run. On basis of the results of this study it is concluded that the ear pulse can be implemented as a feedback parameter for PLL onset. In order to implement the ear pulse waveform as a standard feedback signal during centrifuge training the reliability of the signal must be evaluated under different G-onset rates and with pilots performing an anti-G straining maneuver. Specially, the delay between the changes in the ear pulse waveform and changes in head level blood pressure must investigated under higher G-onset rates. So, the ear pulse wave form changes will have to be correlated with continuously measured blood pressure.

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