4 Chapter 4 - Project: Detection of heart rate by measuring impedance of the wrist

4.1 Introduction

Heart rate is an important physiological measurement for informing the training of athletes. There are a number of techniques available to measure heart rate but, demands for reliable measurement during strenuous exercise places practical constraints on the size and positioning of measurement apparatus and the method used. Devices are commercially available, usually in the form of a wrist watch encompassing several other functions. These are based on measuring an ECG signal, achieved most commonly by supplementing the watch with a remote chest band which is able to make electrical contact across the chest and transmit data to the watch wirelessly (fig 4.1 left). Alternatively, some devices utilise a touch button to establish an electrical connection across the body negating the need for a breast strap (fig 4.1 right). Of course, this can not provide a continuous reading.

Figure 4.1 - Left: Omron HR-100C Heart Rate Monitor (photo from amazon.com)
Right: Reebok Strapless II Heart Rate Monitor Watch (photo from yourhealthcareonline.com)

Neither of these technologies provide the desired solution of a continuous measurement from a single device worn like a watch. This project investigates the practicalities of measuring heart rate by detection of wrist impedance changes to pulsatile blood flow instead of detecting an ECG. In principle this would allow continuous measurement of heart rate from a self-contained device worn at the wrist.

4.2 Background Theory

Impedance plethysmography is a well established field within physiological measurement. It has found widespread use in health care, for example, monitoring of respiration rate from the impedance of the chest. However, it is not commonly used to measure the performance of the heart.

When the heart beats it sends a pulse of high blood pressure (systole) around the arteries. Due to the compliance of the arteries they expand slightly in response. Blood has a relatively high conductance compared with that of other tissues (Eg muscle 0.32S/m, fat 0.02S/m, blood 1S/m @1kHz [Gabriel et al 1996a]). Therefore, it follows that the expansion of an artery at systole would produce a concurrent increase in overall conductance in the region through which it passes.

Two main arteries supply blood to the hand through the wrist; the Radial (the usual site for detecting heart rate by palpitation) and the Ulnar (fig 4.2). Therefore, a change in resistance at the wrist would be expected in response to pulsatile blood flow through these vessels.

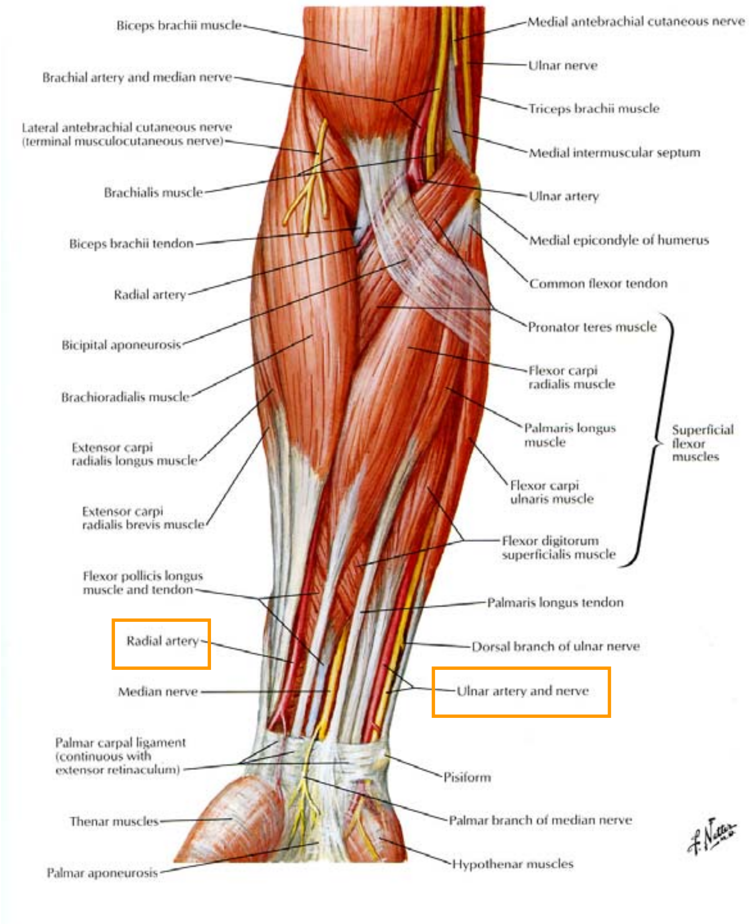

Figure 4.2 – Anatomy of the lower arm (Netter 2006).

Measurement of resistance is theoretically simple to achieve. Using Ohm's law the potential measured in response to a constant current passed through the tissue is proportional to its impedance. Unfortunately, this direct two electrode method is not very sensitive to the tissue changes due to the relatively high (and variable) resistance of the electrode skin interface.

A system which is independent of the electrode tissue interface can be constructed using a tetra polar electrode arrangement. A constant current is passed through the tissue between two dedicated electrodes and a potential difference is measured from a separate electrode pair placed between the current injectors. Negligible current flows through the sensing electrodes when connected to a high impedance amplifier so the potential will not be sensitive to the resistance of the electrode interface. Similarly the constant current drive produces a voltage differential in the tissue which is independent of electrode impedance.

4.3 Existing Research

Farag et al (1994) successfully detected heart rate and respiration rate from the wrist. Their experimental setup involved four electrodes placed around the circumference of the wrist. The current injecting electrodes were at maximal distance either side of the wrist bulk and the sensing electrodes placed in between these. They were successful in detecting impedance signals and went on to investigate the affect of electrode spacing on signal size. Wang et al (2007) used a similar technique when investigating changes in cardiac output. However, they used electrodes spaced out down the whole of the lower arm.

Both studies measured bulk impedance changes in the wrist and did not seek to identify the source of the impedance changes. As the exact source of the phenomena is unknown, a method of investigating the signal at localised areas of the wrist over and away from the arteries supplying the hand would be useful. The sensitivity of a tetra polar device to impedance changes in voxels at various spatial positions bellow the skin surface can be exploited to get the best signal by adjusting the spacing of the electrodes (Brown et al 2000).

The present study aims to verify that bulk impedance changes can be detected and investigate the use of small local electrode arrangements for the detection of local changes in impedance. It also records the magnitude of various movement artefacts in an attempt to assess the viability of using this principle in sports measurement.

4.4 Experimental Methods

Fig 4.3 shows the experimental setup used in this study. A tetra-polar measurement system was constructed using a 5kHz battery powered current source with a pk-pk current of 100uA. A differential amplifier with an isolated front end (Neurostar M592B) was used to amplify the voltage between the detecting electrodes by 54dB. A band pass filter centred at 5kHz (24dB/Octave) was used to isolate the 5kHz carrier before demodulation through an RMS to DC converter (*Analog Devices AD536A*). Finally the signal was band pass filtered from 0.8Hz to 5.8Hz before being viewed on a digital storage oscilloscope.

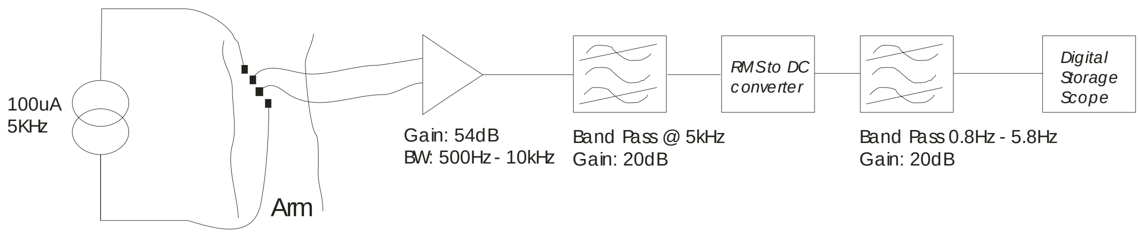

Figure 4.3 – Experimental setup.

For investigation of bulk impedance changes ring electrodes 1cm wide where formed from lead foil and held in place by Micropore tape (fig 4.4). The skin around the wrist was cleaned with an alcohol wipe before applying the electrodes and electrode gel was also used under the lead foil electrodes to decrease electrode impedance.

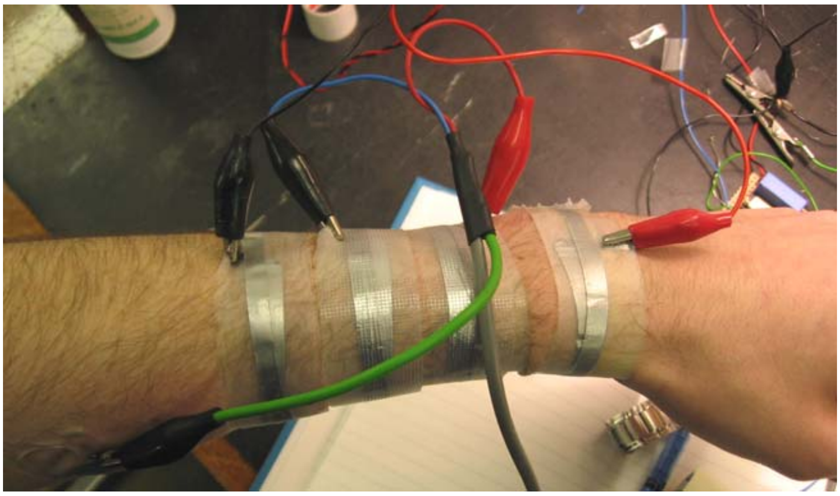

Figure 4.4 - Ring electrode arrangement used for measurement of bulk impedance changes.

To investigate local impedance changes an electrode array was constructed from a row of four 3M bolts 8mm apart secured to a rigid plastic strip (fig 4.5). A separate ground electrode was constructed from lead foil and Micropore tape.

Figure 4.5 - Electrode probe used for measurement of local impedance changes.

For validation purposes the ECG was also detected using two lead foil electrodes placed on the chest, connected to an isolated amplifier and viewed concurrently on the storage scope.

A system for providing a quantification of force applied to the electrode array when placed on the wrist was devised. For early experiments a micro-manipulator was used in a rigid stand to measure compression in mm. For later experiments this was upgraded to a force measurement system constructed using kitchen scales between the micro manipulator and electrode array (fig

4.6). An estimate of pressure could be achieved by associating the weight scale in kg to force in Newton's and then to pressure using the known surface area of the electrode array as a constant.

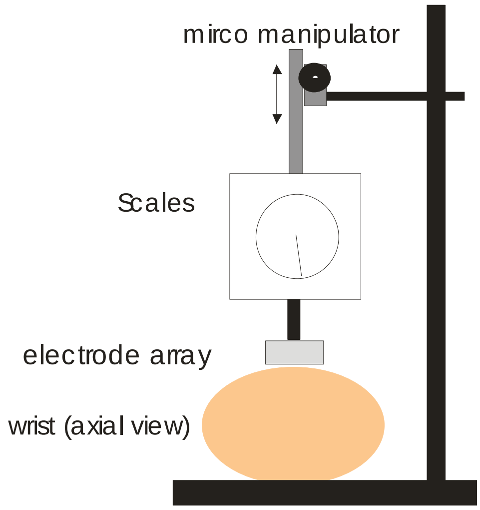

Figure 4.6 – Apparatus used to estimate electrode pressure.

Experiments were initially carried out on my own wrist. Basic waveforms using the local strip electrodes were also collected from three other staff volunteers.

4.5 Results

4.5.1 Ring electrodes

Fig 4.7 shows an impedance signal measured using the ring electrode array (fig 4.4). The signal amplitude is typically ~8mVp-p which corresponds to an impedance change of approximately 0.2% (200mΩ). An ECG signal was recorded concurrently and is displayed with the impedance waveform.

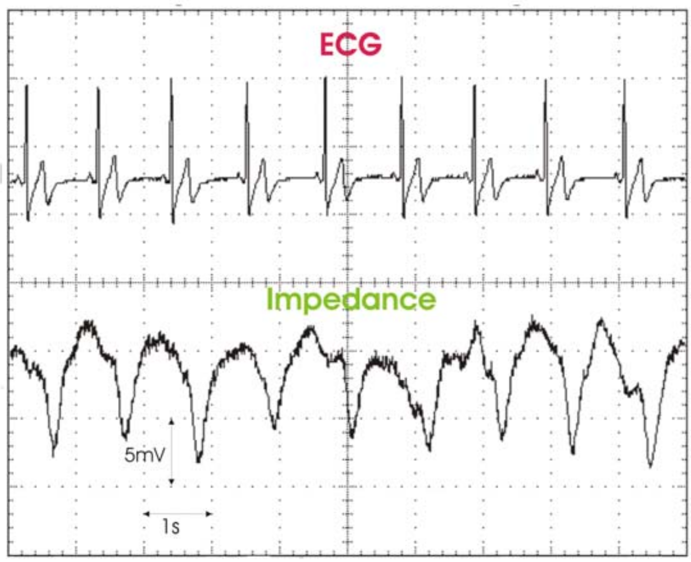

Figure 4.7 - Concurrent recording of wrist impedance and ECG.

4.5.2 Affect of varying drive current frequency

Fig 4.8 shows impedance waveforms for a range of drive current frequencies.

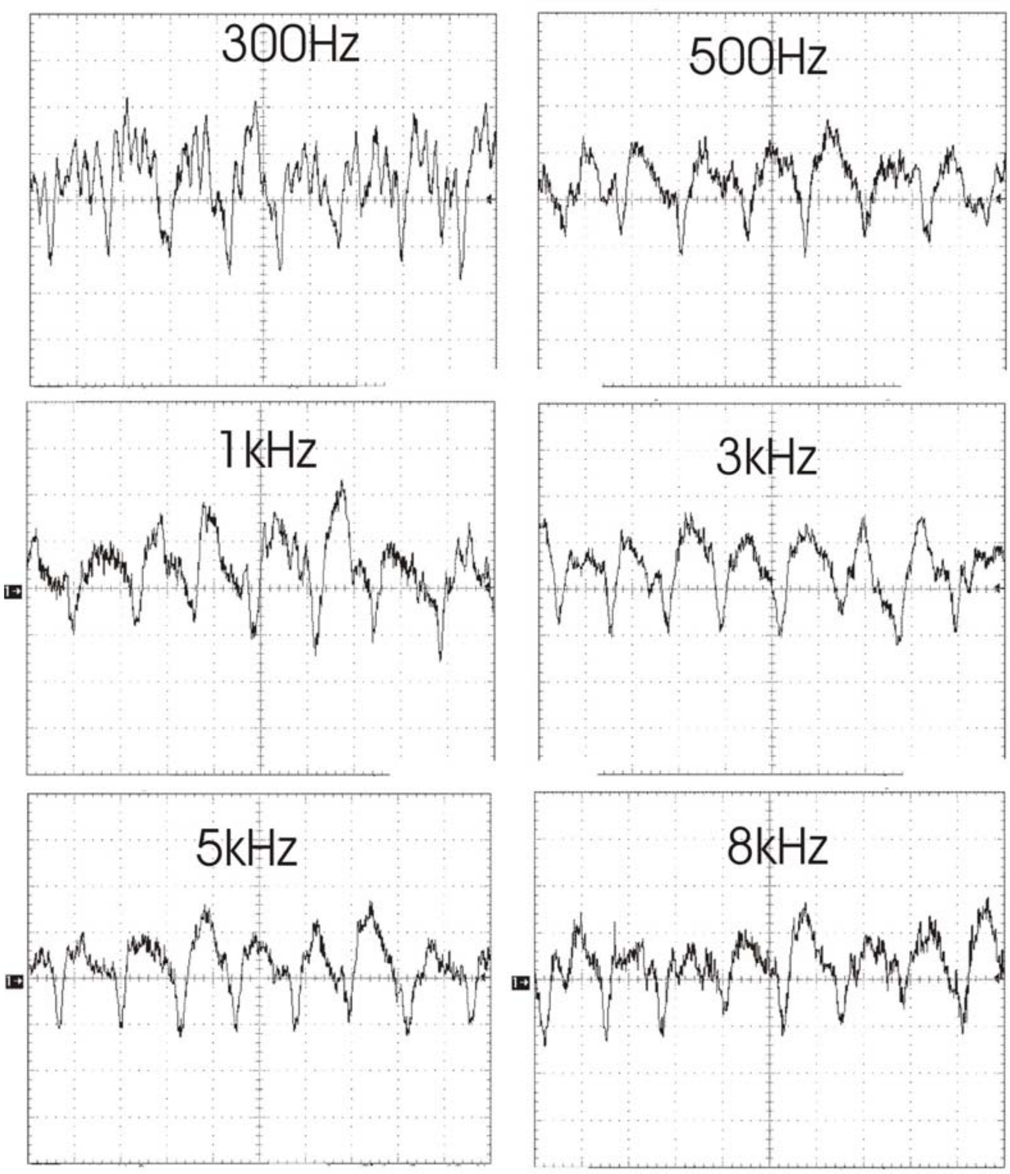

Figure 4.8 - Wrist impedance waveform shapes with current source and first band pass filter set to different frequencies using ring electrodes. Time: 1s/division. Amplitude: 2mV/division for all traces. The first bandpass filter settings were adjusted to remain centred on the drive frequency.

4.5.3 Sensitivity to movement artifact

All signals presented thus far were recorded at rest on the left wrist with a completely relaxed arm and hand. In an attempt to quantify sensitivity to movement artifact signals were recorded in response to a number of maneuvers.

Movement	Signal amplitude
Opposite hand and arm	No affect
Neck	No affect
Left index finger	120mV
Whole left hand	600mV

4.5.4 Electrode array measurements

Fig 4.9 shows the impedance waveform detected using the strip electrode array over the radial artery for myself and four staff volunteers.

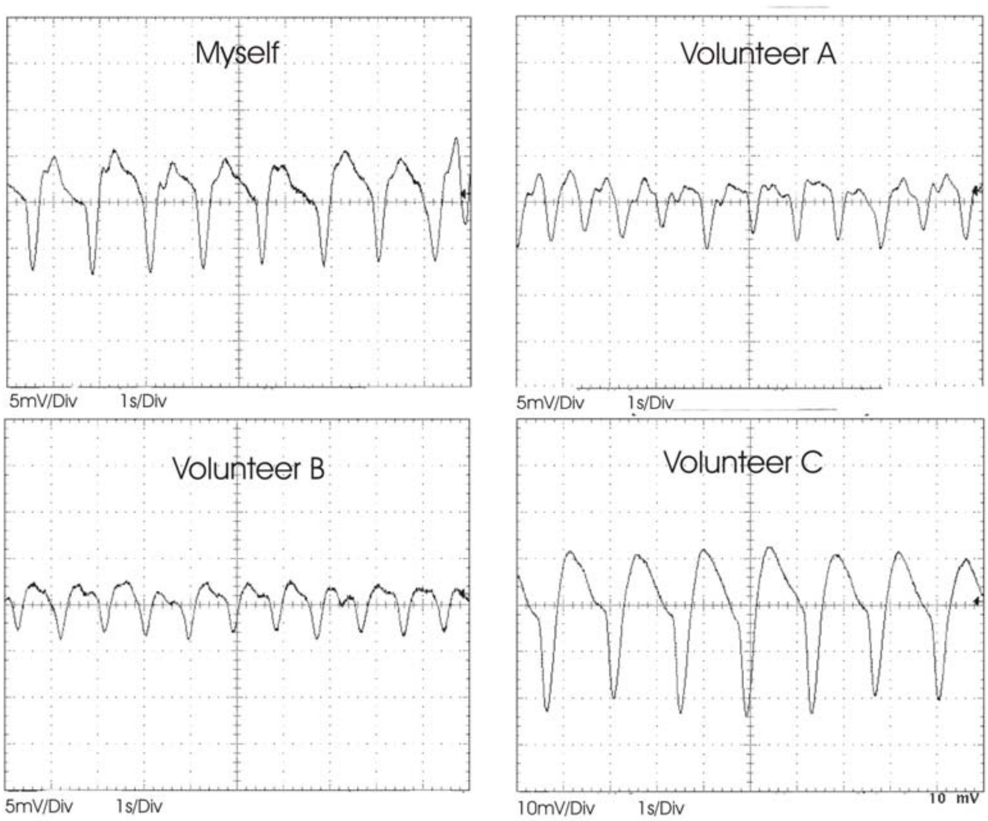

Figure 4.9 - Wrist impedance waveforms using electrode array 6cm up from the wrist over radial artery for four volunteers.

Fig 4.10 shows the variation of signal amplitude with compression of the electrode array onto the wrist over the radial artery for five measurement runs on myself. The force applied at the point of maximum signal (corresponding to a compression of approximately 5mm) was later measured to correspond to a force of 6N.

Surface area of electrode array: $3.5\text{cm} \times 2\text{cm} = 7\text{x}10^{-4}\text{m}^2$

$1\text{Pascal} = 1\text{N/m}^2$ so

$\text{Pressure} = 6\text{N}/7\text{x}10^{-4} = 8.6\text{kPa}$

$= 8.6 \times 760/101.33 = 65\text{mmHg} \pm 10\text{mmHg}$

This is only a crude pressure measurement system. It gives a very approximate but useful estimate of pressure.

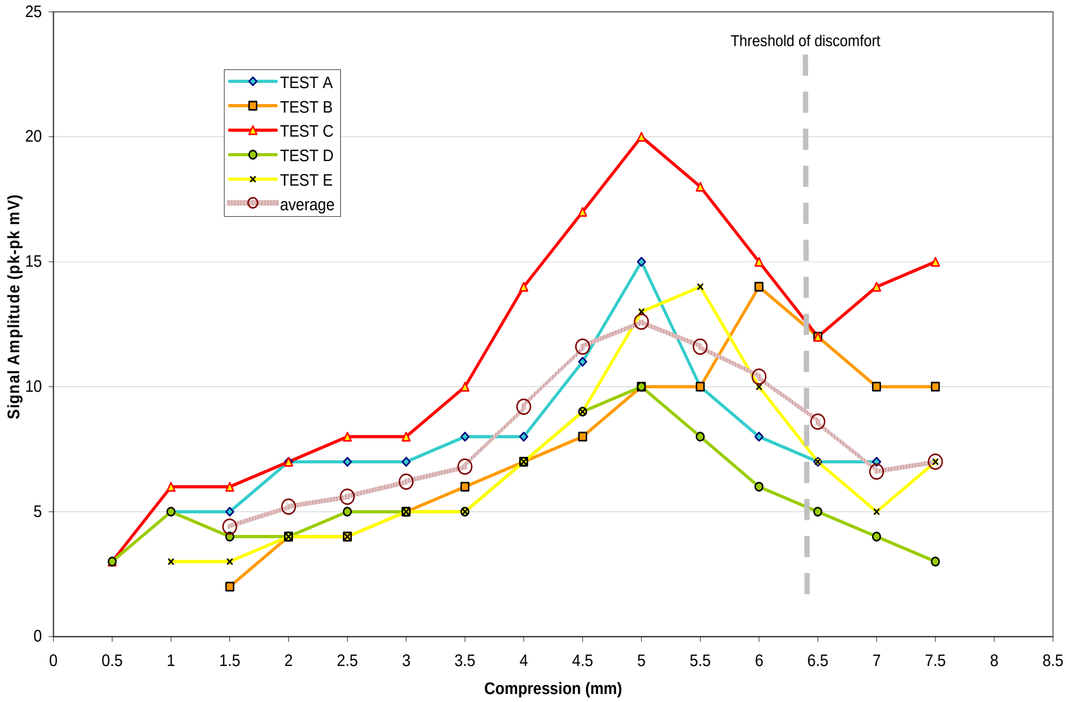

Figure 4.10 - Signal amplitude verses compression of electrode array into wrist over radial artery in the same position. Showing results of five trials and the average.

Fig 4.11-A shows how the signal amplitude varies with position on the wrist. Six readings were taken at each of five positions of the array (red markers), all parallel to the arm length and spaced 1cm apart, and then averaged (blue line). Fig 4.11-B shows the approximate corresponding location of the electrode array on the actual wrist. Fig 4.11C is a sectioned anatomical reference drawing showing the two main arteries that supply the hand.

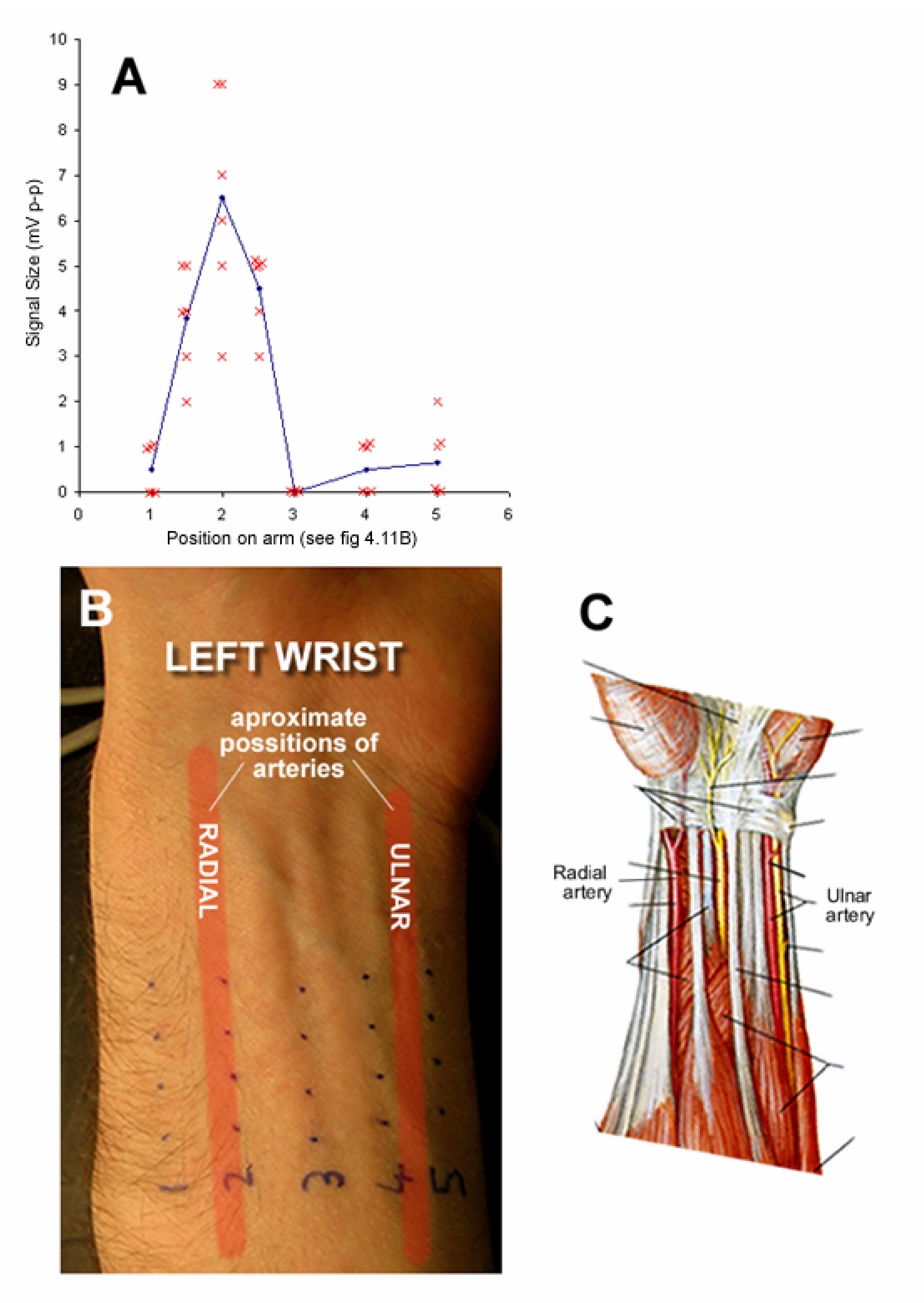

Figure 4.11 – A: Signal amplitude verses electrode position over the wrist. B: Annotated photo of electrode position markers on wrist. C: Anatomical illustration of the wrist.

4.6 Discussion

Good correlation can be seen between the minima of each impedance waveform and the QRS complex of the ECG signal in fig 4.7. There appears to be a phase offset of approximately 0.4s arising from the transit time of the systolic pulse from the heart to the wrist. This recording demonstrates that the signal arises due to pulsatile blood flow and hence can be used as a measurement of heart rate.

Due to the small size of the impedance changes seen (~0.2%), the system has to be very sensitive (total gain of 94dB). Unfortunately due to the relatively high impedance changes resulting from movement artefact, relatively low noise waveforms such as those in fig 4.7 can only be recorded whilst the subject is completely motionless. The amplitudes of signals recorded in response to various trial movements are up to 100 times larger than the signals of interest. This would be a major obstacle to the use of this kind of system in sport which would require heart rate measurement under highly active physical conditions.

Variation of the AC current frequency does not make a noticeable difference to the size or morphology of the signals over the frequency range tested (300Hz to 8kHz). The bandwidth tested was restrained by the high frequency cut off of the isolated amplifier and to avoid 50Hz interference. An unusual unexplained oscillatory interference (about 4Hz) was observed when using a 300Hz AC drive signal (fig 4.8). The drive frequency trials were also performed using the electrode array and were found to give similar results to those using the ring electrodes.

The conductivity of tissue changes with frequency (Gabriel et al 1996a) and the ratios of conductivity of blood to muscle decreases with increasing frequency (Gabriel et al 1996b). Whilst only a limited frequency range has been explored there is no reason to assume sensitivity would increase at higher frequencies.

The signals collected from three other volunteers, although a very limited data set, provided two interesting conclusions. First, heart rate signals can be reliably picked up from others and the waveform shapes were very similar. Secondly, there is a difference in signal amplitudes as is very evident from volunteer C whose signal was over 30mV compared to 10mV or less for the others. A possible explanation for this would be the athletic background of volunteer C, resulting in a higher than average resting stroke volume (Wilmore and Costill 2005). Volunteer C had a waveform which was almost noise free compared with the others but this may have been due to the relatively higher signal amplitude. The differences in thickness of fat around the wrist of each volunteer may also have contributed to the variation of signal amplitudes. Increasing the electrode spacing in the more ‘well built’ volunteers may have provided a increased signal as, in theory, this would increase the depth of sensitivity. There was insufficient time to try this.

The results of pressure verses signal provide convincing evidence of an optimal pressure which appears to be 60-70mmHg for myself. An explanation for the peak in sensitivity at this pressure could be the improving electrode contact, increasing proximity to the radial artery and/or displacement of overlying tissue. As it is squeezed the signal peaks before the pressure begins to cut off the blood flow thus lowering the signal.

The results of the positional trials performed at a constant pressure of 65mmHg are consistent with the anatomical locations of the radial and ulnar arteries. The largest signal was found consistently over the radial artery, perhaps due to its more superficial location. It was not possible to record a signal directly over the tendons in the wrist.

A common feature of all the experiments was the high variability in signal amplitude between trials as can be seen in the compression trial of fig 4.10. This suggests that the signal is highly sensitive to even small shifts in electrode position.

Further investigation could be performed into decreasing the movement artefact. Possible avenues of investigation include, use of more secure electrode arrangements, investigation of alternative electrode types and use of digital filtering to provide greater rejection of noise. However, the experience gained on this project would suggest that such efforts are likely to be unfruitful due to the magnitude of the movement artefacts.

Since this project was completed Gonzalez-Landaeta et al (2008) published similar work detecting heart rate by measuring the impedance of the lower limbs. Their 18 subjects each stood bare footed on a floor pad into which the four electrodes were integrated. Using a very similar detection circuit to ours they reported impedance changes of less than 500mΩ at 10kHz in 18 subjects. Their waveform shapes were also very similar to those in our study as can be seen in fig 4.12. It should be noted that they have inverted their bioimpedance signal in this figure.

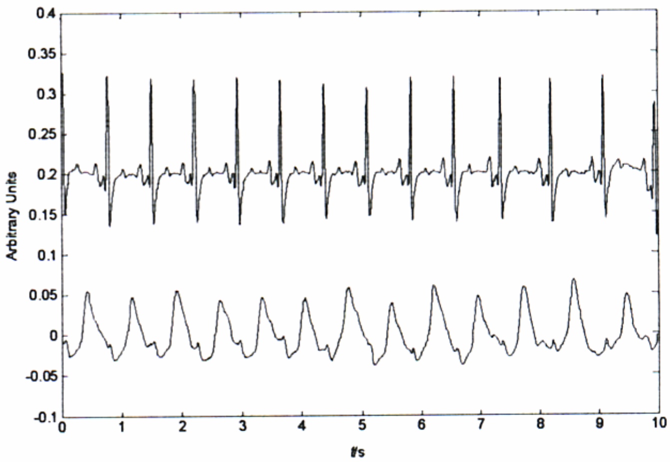

Figure 4.12 – Sample plantar bioimpedance signals (bottom) and concurrent ECG (top) from Gonzalez-Landaeta et al (2008).

4.7 Conclusion

Heart rate signals can be detected from the wrist by measuring impedance. Both ring electrodes and a localised electrode arrays can reliably detect a signal. The highest signals were detected directly over the radial artery when the electrode was applied with a pressure of some 65mmHg. However, the magnitude of the signals was highly dependent on electrode position.

Signal artefacts due to movement were up to 100 times larger than the heart rate signals and as such preclude the use of this technique in active sports applications. It is unlikely that sufficient filtering of this artefact could be achieved and therefore using this technique in sports science is probably not feasible.

4.8 References

Brown, B.H. Wilson, A.J. Bertemes-Filho, P. 2000. Bipolar and tetrapolar transfer impedance measurements from volumeconductor. Electronics Letters. Volume: 36, Issue: 25.

Farag AA, Tacker WA, Foster KS, Bourland JD, Geddes LA. 1994. Detection of pulse and respiratory signals from the wrist using dry electrodes. Biomed Instrum Technol. Jul-Aug;28(4):311-4.

Gabriel C, Gabriel S, Corthout E. 1996a. The dielectric properties of biological tissues: I. Literature survey. Phys Med Biol. Nov;41(11):2231-49.

Gabriel C, Gabriel S, Corthout E. 1996b. The dielectric properties of biological tissues: III. Parametric models for the dielectric spectrum of tissues. Phys Med Biol. Nov;41(11): 2271–2293.

Gonzalez-Landaeta R, Casas O, Pallàs-Areny R. 2008. Heart rate detection from plantar bioimpedance measurements. IEEE Trans Biomed Eng. Mar;55(3):1163-7.

Netter F.H., 2006, Atlas of human anatomy. 4th ed. Pennsylvania: Saunders.

Wang JJ, Wang PW, Liu CP, Lin SK, Hu WC, Kao T. 2007. Evaluation of changes in cardiac output from the electrical impedance waveform in the forearm. Physiol Meas. 2007 Sep;28(9):989-99. Epub 2007 Aug 21.

Wilmore JH and Costill DL. (2005) Physiology of Sport and Exercise: 3rd Edition. Champaign, IL: Human Kinetics