Force Plate Measurement of Human Hemodynamics

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Abstract

We show that force plate measurements provide a noninvasive method to display the motion of the heart muscle and the subsequent propagation of the pulse wave along aorta and its branches.

I Introduction

The aim of this paper is to present a new method to handle multivariate time series obtained by force plate measurements. The proposed technique is subsequently used to display marks of cardiac activity.

We understand the measured time series as coordinate projections of a multidimensional curve. To extract information contained in the signal we will investigate the geometric properties of this curve. In particular we focus on invariants describing the curve as a geometrical complex. Such invariants do not change when the curve is rotated, translated or its parametrization is changed. This is in contrast with the coordinate projections (measured signals) that change under such transformations. A measured signal changes if we modify for instance the allocation of the electrodes during electrocardiographic measurement (ECG). The recorded process, i.e. the electric activity of the heart, however, does not depend on the particular position of the electrodes. So the main hypothesis presented here is that a process is adequately reflected by the total curve and not merely by its projections i.e. by the particularly measured signals.

The geometrical invariants used to describe a multidimensional curve are related to the Cartan - Frenet - Serret concept of moving frames and are known as Cartan curvatures. (Similar methods are used for instance to describe the space time geometry in the general relativity theory of Einstein.) We will demonstrate that the processes causing changes in the measured signal manifest themselves as rapid changes (peaks) in the Cartan curvatures.

Force plate is a device that measures the components of a force and of the related force moments acting on a specified platform. It exploits four sets of transducers located in the four corners of the plate. Since the force vector is, by definition, perpendicular to its moment, only five of the six measured quantities are in fact independent. Hence a measurement of the load dynamics leads to five independent time series. They are usually represented by the three force components together with the point where this force acts on the platform. The point of application of the force provides only two-dimensional information since it is bound to the upper platform of the force plate. This point can be easily calculated from measured moments and forces.

In medicine the plate is usually used to measure forces exchanged between the subject’s feet and the floor (ground reaction). Thus, it can be used for various tests, such as a study.
of maintaining of human balance during quiet upright stance, weight lifting analysis, stability and co-ordination tests or gait analysis.

Our aim here is to demonstrate the facilities of the geometric signal analysis. This is why we focus on a subtle process, namely on the force manifestation of the cardiac activity and hemodynamics. Cardiac activity affects of course the balance and its interference with the postural sway was studied for instance in [2, 3].

To eliminate the interference of the cardiac activity with postural changes and body tremor we measured the subjects in a horizontal position lying on a special bed, cf. [4].

For a reclining subject the motion of the internal masses within the body has a crucial effect on the displacement of the center of gravity (COG) of the body and on the ground reactions of the plate. When a bed with the lying subject is placed onto the force plate, measured ground reaction forces contain information on the blood mass transient flow at each heartbeat and on the movement of the heart itself. (There are also other sources of the internal mass motion that cannot be suppressed, like the stomach activity etc, but they are much slower and do not display a periodic-like pattern.)

Rapid heart contract or expand movements such as e.g. isovolumetric contraction or valve opening are not associated with a significant reorganization of the internal mass. Hence they lead - due to the action/reaction principle - to rapid changes of the ground reaction force without large changes of COG position. The motion of the heart muscle and the related blood flow imply on the contrary an significant internal mass motion. The resulting changes of the ground reaction force are in this case accompanied with substantial changes of the force moments and hence of COG.

The blood flow is associated with a pulse wave that runs mainly on the large elastic arteries (aorta and its main branches) and interacts with the vessel branchings. At a branching, the incoming pulse wave is partially reflected and partially divided into the (two or more) branched vessels. Such interaction between the pulse wave and the vessel branching leads again to changes in the measured signal.

The study of the motion of the body related to the heart activity is usually called ballistocardiography. Discovered in the 19th century, ballistocardiography was studied extensively from the 1930s until the late 1960s, see e.g. [3]. Most of the early ballistocardiographs (BCG) tracked the movement on a special bed on which the patient lay. Later, various accelerometers attached directly to various places of the body (most often to the sternum) were used for the BCG acquisition. However, these direct methods survey only one-dimensional accelerations.

The cardiac activity has a quasiperiodic character and is triggered by an electric signal that can be monitored by ECG. Simultaneous measurement of the ECG signal and of the subsequent mechanical reaction of the cardiovascular system provide interesting information. For instance estimating the time delay between the opening of the aortal valve and the arrival of the pulse wave to the mesenteric artery branching provide an information about the pulse wave velocity and hence about the elasticity of the aorta. An unexpected pulse wave reflection can give a hint on a presence of stenoses or aneurism etc.

Using geometrical invariants of the signal curve we can identify significant mechanical events like valve opening, arriving of the pulse wave to the abdominal aorta branchings etc. The proposed method has been tested on a group of volunteers. The obtained results are reproducible, i.e. they lead to the same results when we repeat the measurement with the same person (although they are slightly different when different people are measured ). It has also to be stressed that only the mechanical response of the force platform and the ECG signal are measured, i.e. the proposed method is fully noninvasive.

II Experiment and data processing

A Bertec force plate, model 4060A, working on the strain gauge technology principle was used. Its length was 0.6 m and its width was 0.4 m. Forces and moments were measured with respect to the following reference system: x-axis parallel to the shorter side of the top plate, y-axis along the
exists unique family (E_k, . . . , E_n) of orthonormal n-dimensional vectors having a positive orientation, so that for every k = 1 ≤ k ≤ n − 1, the kth derivative e^(k)(t) is a linear combination of (E_1, . . . , E_k) for every t ∈ I. The family (E_1(t), . . . , E_n(t)) is called the distinguished Frenet frame. The curvatures κ_i, i = 1, . . . , n − 1 of the curve c are defined using Frenet-Serret formulæ as
\[ κ_i(t) = \frac{E_i'(t) \cdot E_{i+1}(t)}{\|c'(t)\|}. \]

Hence the curvatures characterize local changes of the coordinate system related with the curve. In our case the signal curve is 6 dimensional and is characterized by 5 curvatures.

A mechanical event like a mass motion inside the body or the scattering of a pulse wave on arterial bifurcations leads to changes in the force plate response. It might be difficult to trace out the response by inspecting directly the time series representing the force and moment projections. It is however reasonable to assume that a presence of such an event will locally change the geometry of the total signal curve and will be visible as a local change of its invariants.

To filter out the information related to the hemodynamics we use the fact that this process is quasiperiodic and triggered by the ECG signal cf. [5]. We take the ECG R wave as the initializing event and regard the hemodynamics as being time locked with it. We will therefore evaluate the curvatures for time periods of 800 ms after each R wave and finally take the mean over all R waves.

A similar method is used when an evoked brain activity is studied. In this case the brain activity is triggered by an external stimulus (for instance a periodic flashing light) and a the brain response
is studied by taking the mean of the EEG signal over all stimuli. The paradigm to be followed
is related to the hypothesis that the intestine hemodynamical events change the direction of the
signal curve and are therefore associated with clear maxima of the averaged curvatures.

III Results

The ejection of blood by ventricular contraction spreads over the elastic central arteries. The
augmentation propagates in a form of a pulse wave along the arteries and converts finally the
pulsatile action of the heart into continuous laminar blood flow. On branching places of large
arteries the pulse wave is scattered and the subsequent elastic recoil contribute to the force changes
measured by the plate. A similar recoil is expected also when the artery changes promptly its
direction (like for instance in the aortal arc).

The main event of this type is the scattering of the pulse wave on the branching of the
abdominal aorta into the two iliac arteries. After this reflection a part of the wave propagates
backwards through the aorta whereas the transmitted wave continues its down movement along
the iliac arteries. When passing a branching point the reflected wave leads to similar (but weaker)
recoil as the original wave. It is partially bounced back, i.e. propagates in the same direction as
the primary wave and so on. This means that the prime measured events are accompanied with
echoes, i.e. with secondary events that are placed symmetrically on the time axis. For instance the
renal artery branching (and narrowing of the aorta at this place) leads to a well measured event.
It is subsequently followed by the event triggered by the iliac bifurcation that leads to a massive
bounce back of the pulse wave. Next it is followed by an echo of the renal artery event evoked
by the bounced wave. Hence the primary renal event and its echo are placed symmetrically with
respect to the iliac bifurcation time. Generally the presence of echoes presents a difficulty by the
final interpretation of the signal. Especially by elderly persons when the elasticity of the artery is
already reduced and the scattering events are more pronounced.

Another fact to be mentioned is the interference between the hemodynamical and cardiac
effects. In particular in the second half of the cardiac cycle the hemodynamics triggered by the
heart contraction pursue to develop (and fades out) whereby the heart muscle start to prepare
itself for another beat (diastole). In such a way the force plate is influenced by both processes
simultaneously and this makes the interpretation of this period difficult. This is why we will
analyze the signal starting at the R wave of the ECG signal and ending at roughly 80% of the
duration of the cardiac cycle (800 ms).

III.1 Interpretation

A typical example of the averaged first four curvatures of the measured 6-dimensional curve is plot-
ted in the figure together with the averaged ECG signal, phonocardiogram and the time behavior
of the reaction force components $F_x$, $F_y$ and $F_z$. The overall pattern of the first four curvatures
appears to be reproducible across the sample of all examined subjects. The fifth curvature could
not be classified because of a high level of noise.

In particular, we have identified 7 peaks in the first curvature that reflect direct hemodynamical
events - see figure. These peaks are visible in all measured subjects and most of them appear
also in the other three curvatures.

The marked peaks can be assigned to particular hemodynamical events as follows.

- Peak 1: appears 50 ms after the R wave together with the onset of the first heart sound.
  This peak reflects the enclosure of the tricuspid and mitral valves. It is followed by the
  isovolumetric contraction of the heart muscle.

- Peak 2: reflects the isovolumetric contraction of the heart muscle. Typically this peak
  appears 75 ms after the R wave.
• Peak 3: This smaller peak reflects the opening of the aortic and pulmonary valves and ventricular contraction. It is accompanied by the increase of the intensity of the first heart sound that is caused by oscillations and turbulency of the ejected blood. The peak appears 250 ms before the onset of the second heart sound and approximately 110 ms after the appearance of the R wave.

• Peak 4: reflects the scattering of the pulse wave on the pulmonary artery bifurcation. It shows up sometimes is form of two peaks due to the recoil of the pulse wave on the complex branchings of the pulmonary artery. The doubling of the pulse waveform has been measured also with the help of pulmonary arterial catheterization. This peaks appear with a time delay of 110-140 ms.

The peaks 3 and 4 become in some cases very close and merge to a single blurred peak.

• Peak 5: reflects the passage of the pulse wave through the aortal arch and is a consequence of the sudden change in the direction of the pulse wave. It is accompanied with a fading of the first heart sound. It succeeds the peak 4 since the ascending part of the aorta is longer then the length of the ascending part of the pulmonary artery. Its typical time delay with respect to the R wave is 170-190 ms.

• Peak 6: Reflect the scattering on the sequence of the abdominal aortal branchings. When the subject reclines on belly the peak is usually more pronounced or even splits into two narrow peaks. Moreover the final abdominal branching (renal branching) is accompanied also with a narrowing of the aorta. Its diameter changes form approximately 2 cm to 1.3 cm and this leads to a well pronounced scattering event. The peak 6 appears typically 280-330 ms after the R wave onset.

• Peak 7: is related to the iliac branching of the aorta and represents the most pronounced pulse wave reflection during the cardiac cycle. It emerges with a time delay of 400 - 480 ms after the R wave of the ECG signal.

All the above described curvature maxima were visible by all measured subject and are directly related to well defined morphological structures. They are, however, followed, by 2 or 3 further curvature peaks that are not of a clear morphological origin. An typical example of this type of peaks are the maxima denoted in the figure 2 as 8 and 9. They appear with time delays exceeding 500 ms. Moreover they form an almost periodic pattern. These peaks are easily explained as echoes of the renal branching reflection. The point is that, as already mentioned, the branching of the abdominal aorta into the common iliac arteries leads to the utmost reflection of the pulse wave. Part of the pulse wave travels back, is again scattered by the renal branching and appears as an echo of the original renal curvature peak (peak 8). The scattering continues by backscattering that travels downwards the aorta. So the next maximum (peak 9) appears as an echo of the original iliac branching (peak 7). In majority of cases one more peak (labelled as 10) is visible representing a second echo of the peak 6. Since peaks 8 and 9 are nothing but echoes of the peaks 6 and 7 respectively the time dependence of the forces measured by the plate has to display a symmetry against the related reflection point. We demonstrate this fact in the figure 2.

A direct insight into the reflection process is obtained by patients with pathological changes on the abdominal aorta. Usually the detailed structure of the aorta is in this case known from the angiographical screening. The pathological changes lead however also to changes in the signal measured by the force plate. In such a way new curvature peaks appear and can be directly ascribed to the pathological aortal structures.

To demonstrate the existence of the above described oscillations we measured a man with abdominal aneurism. The aneurism was localized between the renal and iliac branchings. The pulse wave reflection on the aneurism leads to the appearance of a sharp curvature peak that is
localized between the standard “renal” and “iliac” peaks. The strong reflection on the aneurism and the consequent backscattering leads also to a rise of additive echo of the renal peak. This echo, however, falls into the same time interval as the peak caused by the iliac branching which means that this peak appears blurred. After the iliac scattering the backward pulse wave hits the aneurism again and is scattered there. An echo of the aneurism scattering is the consequence. The distance between the iliac branching and the aneurism is smaller then the distance to the renal branching. Hence the aneurism echo appears in shorter times then the renal echo by healthy people. The subsequent oscillations display therefore a much shorter period then by healthy subjects. We have compared the results of the patient with a result obtained by healthy volunteer of the same height and displayed them in the figure 4.

We have measured subjects also lying on the belly. The results are very similar to those with subjects resting on their back. The only difference is that usually the curvature peaks related to the abdominal aorta branchings are more expressed. Also peaks related to the motion of the heart muscle are more explicit. It is related to the fact that lying on the belly the chest of the subject is in direct contact with the bed and the heart motion is better transferred to the force plate.

IV Discussion

First we have to comment on the fact that the height of the curvature peaks is not directly related to the strength of the reflection. Curvature is an abstract and purely geometric characterization of the process. Its peaks reflects changes of the local reference system bound to the signal curve. Hence they provide information about the times the inherent processes starts and ends (sharp peaks are related to quick changes). But they do not give a direct reference on the process strength.

Aortic pulse wave velocity (PWV) is widely accepted as an index of aortic stiffness with the application of predicting risk of heart disease in individuals [6]. Since aorta is not directly accessible various methods are used to obtain the value of PWV on various aortic parts. Some of them are invasive using catheters with micromanometric sensors [7]. Other methods are based on high sampling rate magnetic resonance imaging (MRI) of various aortal segments [8, 9]. The curvature analysis of the force plate data offers another way to obtain aortic PWV.

Knowing the reflection times $T_1$ and $T_2$ of two events and the approximate distance $L$ between the two reflection points, PWV can be easily established as $v = L/(T_2 - T_1)$. The total length $L$ of a human aorta (ascending aorta + aortic arch + descending aorta up to the iliac branching) is $L \approx 70 \text{ cm}$ [10]. The two relevant events used are the opening of the aortic valve (peak No 3) and the iliac branching reflection (peak No. 7). The obtained PWV is equal to approximately 3.9 m/s and is in full agreement with the values obtained by the MRI and manometric measurements.

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References


Figure 1: Averaged ECG, phonocardiogram, four curvatures, force components $F_x$, $F_y$, $F_z$ and coordinates of COP for a reclining subject.
Figure 2: Averaged ECG and first curvature for a reclining subject during 750 milliseconds after the onset of the ECG R wave. The upper part of the figure shows the average curvature, lower part displays the first curvature during separate heart beatings.
Figure 3: The lower part of the figure displays the time dependence of the first curvature. The peaks are labelled with numbers 6 – 10. Numbers referring to original peaks are in a box, the related echoes are in a dashed box. The upper part of the figure displays the curvature as a function of the second force component. The oscillatory behavior of the force is clearly visible.
Figure 4: The lower part of the figure displays the first curvature obtained for a patient with abdominal aneurism. The upper part shows the data obtained for a healthy volunteer of the same height. The significant curvature maxima as well as the period of the oscillations are marked by arrows.