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Hemodynamics as a possible internal mechanical disturbance to balance

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Abstract

The postural control system is assessed by observing body sway while the subject involved aims at maintaining a specified up-right posture. Internal masses generate internal reaction forces that constitute an internal mechanical stimulus that may contribute to cause segmental displacements, i.e. body sway. Thus, gaining knowledge about the amplitude and direction of these reaction forces would contribute to gain insights into the mechanisms that influence the maintenance of balance and into its control. The 3-D force vector that acts on the body centre of mass (COM) and is associated with the transient blood movement at each cardiac cycle was assessed in a population sample of 20 young adults during the maintenance of a quiet up-right posture. Typical patterns of the three components of this force vector were identified. Relevant parameters were selected and submitted to sample statistics. For a number of them, linear correlation with subject-specific parameters was found. The antero-posterior force component was characterised by a triphasic major wave, the peaks of which had values up to 0.40 N. The vertical component showed a repeatable triphasic wave with peak-to-peak values in the range 1.3–3.0 N. The medio-lateral component showed relatively low peak-to-peak values (in the range 0.05–0.10 N). The resultant vector had an amplitude that underwent several oscillations during the cardiac cycle and reached its maximal value in the range 0.6–1.7 N.

Keywords: Posture; Hemodynamics; Force plate

1. Introduction

The performance of the postural control system may be evaluated by observing and analysing body sway while the subject involved aims at maintaining a specified up-right posture. The analysis may be based on the measured movements of selected ensembles of body segments, looked upon as if they were rigid bodies, and, seeking a more compact approach, on the reconstructed trajectory of the centre of mass (COM) [1,2]. Also the forces exchanged between the subject's feet and the floor (ground reactions) contain information on body sway. The relevant resultant force is proportional to the acceleration of the whole body COM. The intercept of this force with the support base, the so-called centre of pressure (COP), depends on both position and acceleration of the COM in addition to the angular acceleration of the moving body portions. Given the oscillatory nature of body sway, associated accelerations, and thus ground reactions, are likely to be more sensitive to physiological or pathological modifications of sway than the displacements as provided by the tracking systems used to record body segment movement. The latter measuring systems, besides suffering from a relatively low signal-to-noise ratio [3,4], provide relevant information through models the structure and parameters of which are intrinsically inaccurate. It must be added that experiments carried out using the dynamo-

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metric platform are straightforward and minimally interfere, either physically or psychologically, with the test subject as opposed to the displacement tracking systems. The above considerations account for, and justify, the reiterated attempts carried out by both researchers and professionals to extract information concerning a specific subject's ability to maintain a given up-right posture from force plate readings only.

Regardless of whether body sway has been looked upon as the output of a multiple input/disturbance system or as a stochastic process, the attempt to formulate hypotheses concerning the control mechanisms underlying the maintenance of the up-right posture has led to considering the human body as being made of a more or less complex chain of rigid links connected by hinges. Internal masses in motion within the body, such as those associated with the cardiovascular system and with the blood mass transient movement at each heartbeat, and with ventilation, have seldom been taken into consideration. These masses in motion contribute, by definition, to the whole-body COM displacement and, thus, to the ground reactions. In addition, they generate internal reaction forces that constitute an internal mechanical stimulus and, as such, a disturbance to the postural control system that, in turn, may cause segmental displacements.

Thus, after subtracting the gravitational force, the ground reactions recorded during the maintenance of an up-right posture may be looked upon as the resultant dynamic action associated with the following kinematic factors:

1. externally observable movements of the body segments, globally referred to as body sway, and
2. movements of internal masses.

Since, when evaluating a person's ability to maintain balance, the focus is on body sway, the share of the ground reactions associated with the movement of internal masses is to be regarded as an artefact (something similar to the stimulus artefact well known in electrophysiology).

As far as ventilation is concerned, studies have been published that state that it may constitute a disturbance to the postural control system and result in a considerable portion of body sway [5–7]. The possibility of suppressing ventilation during posturographic experiments allows for a quantitative assessment of the related effect [6]. However, most authors reporting on posturographic results do not mention this aspect or take it into account.

To our knowledge, hemodynamic effects on the maintenance of balance and on the variables used for subject postural ability assessment have been hypothesised but not described [8–12]. Bircher et al. [9] and Stürm et al. [10] have speculated on the ground reaction force components and on the relevant rectified impulse as measures of what they called whole-body endogenous microvibrations, during quiet standing. Although they emphasised the fact that the cardiac activity was somehow embedded in those force signals, with special reference to the vertical force component, they did not isolate the relevant internal reaction force or tackle the problem of its effect on the maintenance of balance.

Thus, concerning the force acting on the whole body COM and associated with cardiac activity and transient blood flow, which will be referred to in the following as cardiac activity force (CAF), the following questions still await an answer.

1. What is the amplitude and direction of the internal reaction forces elicited by hemodynamics (CAF)?
2. What is the contribution of blood movement alone to the entire body COM and to the COP trajectories?
3. Is this hemodynamic perturbation and its mechanical effect on the entire body quantitatively relevant within the process of balance control?
4. Does this internal mechanical stimulus induce an artefact that may significantly affect the value of the parameters normally used to evaluate a given subject's postural stability?

Answering these questions would entail contributing to add insight into the mechanisms that influence the maintenance of balance and into its control, and would help to improve the experimental and analytical techniques aimed at assessing a person's ability to carry out this motor task.

This paper deals with the first question. The CAF vector, as assessed in a population sample of young adults during the maintenance of a quiet up-right posture, is presented and its features discussed and generalised.

2. Materials and methods

2.1. Experimental set-up

A six component force plate was used (Bertec Corp., mod. 4060 A). Its dimensions were 0.6×0.4 m and it carried resistive strain gage load cells located at each of the four corners. Strain gages were wired so that six Wheatstone bridges, one for each load component, were constructed. Forces and moments were measured with respect to a reference system consistent with the International Society of Biomechanics convention [13], x-axis parallel to the long side of the top plate, y-axis vertical upwards, and z-axis along the short side of the top plate. Relevant electrical outputs were preamplified within the force plate and then amplified by a selectable gain in an external unit. The amplified force plate signals were fed to a low noise acquisition system (Step PC®, by Medway SA, Switzerland) that included a 12 bit analogue to digital converter. Gain and offset of the individual channels of the measuring chain were set so that, during the experiments, recorded signals had a mean value close to zero and most of the dynamic range of the A/D converter could be exploited. A hardware filter limited the overall bandwidth of the six force plate channels to 450 Hz. Electrocardiographic (ECG) signals (II lead) were also recorded using the same acquisition system. Data were acquired at a sampling rate of 2000 samples per second for an adequate temporal resolution and avoidance of the filter roll off effect.

2.2. Experimental protocol

A sample of 20 healthy young adults (ten males and ten females, aged 22–40, body mass 47–83 kg, stature 1.63–1.81 m) was investigated after informed consent had been obtained.

Subjects were asked to step on the central region of the force plate facing the positive direction of the x-axis and to assume the following up-right posture: arms lying alongside legs; lateral malleoli distance equal to iliac spine distance; eyes open and looking straight ahead at a 3 m distant visual reference. Ambient illumination and noise were kept under control. Subjects were instructed to maintain a still, even if not rigid, stance and breathe normally. Data acquisition started 10 s after an operator command to the subject and lasted 90 s. Immediately prior to the subject stepping on the force plate, the output of each channel was acquired for 5 s and the mean value calculated thus obtaining an estimate of the channel offsets.

A sequence of three trials was carried out at 10-min intervals for each subject. The sequence was then repeated 1–3 days later.

2.3. Digital signal processing

After acquisition, the relevant channel offset values were subtracted from the dynamometer signals. These were multiplied by the force plate calibration matrix that included the gain factors. In this way, the crosstalk between channels was compensated for, and the raw data were converted into force and moment components expressed in SI units and referred to the above-mentioned system of orthogonal axes. The force components of interest in the present study were then filtered using a FIR low pass filter with an attenuation slope of 30 dB/oct and a cut-off frequency of 25 Hz. The latter frequency value was a conservative choice based on previous frequency analyses of the signals involved [14,15]. When the force plate was loaded with a dead weight and signals were acquired and processed as illustrated above, the three force component variations relative to the respective mean values consisted of noise added to a low frequency drift. The r.m.s. values of the noise and the drift turned out to be 0.03 and 0.008 N for the medio-lateral component (quantisation error = 0.003 N), 0.05 and 0.01 N for the antero-posterior component (quantisation error = 0.003 N), and 0.07 and 0.01 N in the vertical direction (quantisation error = 0.006 N).

Each recording session yielded simultaneous measurement samples of cardiac electrical activity and of the three components of the ground reaction force. For each trial, the samples corresponding to the R waves of the ECG were identified and used to segment the force signals. The segments were of different duration due to cardiac variability.

Each segment of each force component was assumed to contain an identical deterministic signal that described a trial specific CAF component time history along an average cardiac cycle. This deterministic signal was superimposed to a number of signals that were assumed to be uncorrelated with respect to cardiac activity. These were: a noise, with zero mean value, produced by the dynamometric sensors and the other elements of the measuring chain; a signal that came from the sum of the force signals associated with body sway, with skeletal muscle tone and internal mass movements different from those associated with the cardiovascular system; an error signal associated with the time varying orientation of the oscillating trunk and, thus, of the relevant portion of the cardiovascular system, relative to the laboratory frame with respect to which the deterministic CAF was estimated; an error signal which accounted for the fact that the CAF time function varied at each cycle because of cardiac variability.

Thus, by coherent averaging of the above-mentioned signal segments, aligned with the corresponding R wave sample, an estimate of the three force signals correlated with the cardiac activity, i.e. of the CAF components, was determined. For each signal, 80–100 epochs were summed, thus obtaining a theoric improvement of about 18–20 dB in the signal-to-noise ratio. A few experiments were carried out that lasted longer (up to 400 epochs) but no significant improvement was noticed in the results.

2.4. Feature extraction and statistical analysis

For each trial, the above-illustrated data processing procedure provided a mean heart frequency (f) together with an estimate of the CAF components (F_x, F_y, F_z) and of the resultant (F) amplitude and orientation time histories. The instantaneous orientation in space of the resultant CAF was described using the three angles (φ_{xy}, φ_{zy}, φ_{zx}), formed by the projections of this vector onto the three laboratory planes xy, zy, and zx, with the x, z, and, again, z-axis, respectively, positive counterclockwise. Time was expressed as a percentage of the mean cardiac cycle (T).

For each subject, the CAF components and resultant amplitude and orientation angle time histories were averaged over the six trials. In this way, a representative CAF versus time pattern and the relevant confidence limits were obtained for each subject.

In order to characterise these time histories, parameters representing selected peak values and the time instants at which these peaks took place were extracted. In addition, the CAF component time histories were rectified and integrated over the mean cardiac cycle, thus obtaining rectified force impulses (I_x, I_y, I_z). This was done after Bircher et al. [9] and Stürm et al. [10]. The mean heart frequency and the above-mentioned parameters, from now on referred to as CAF parameters, for all trials, represented the data set on which statistical analysis was performed.

In order to evaluate the robustness of data in terms of short-term test–retest reliability, the intra-class correlation coefficient (ICC) together with a repeated measures analysis of variance was used [16,17].

Both within- (six trials) and between-subject (20 subjects) descriptive statistics were performed and the following information was obtained for each parameter: sample mean value (m), standard deviation (S.D.), and coefficient of variation ($CV\% = S.D./m \times 100$).

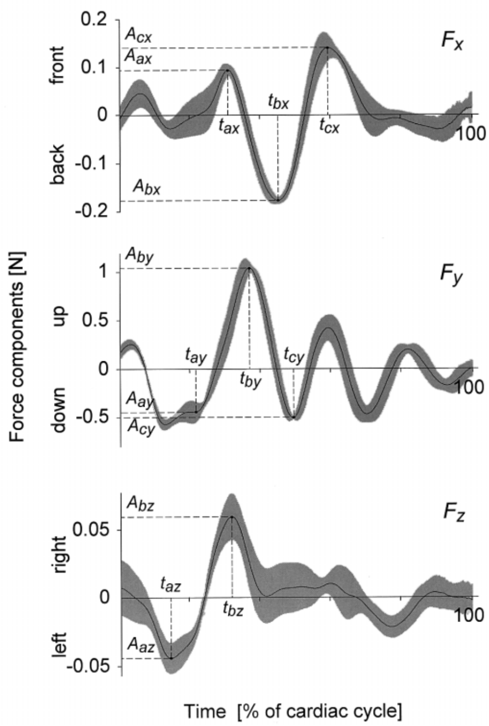

Fig. 1. CAF components with 95% confidence limits, during a cardiac cycle, as obtained in a female subject (36-year old; height, 1.63 m; body mass, 56 kg; mean cardiac cycle $T = 0.56$ s). Peaks, with an amplitude and occurrence time included in the CAF parameters (see Table 1), are also indicated.

In both a correlational and predictive perspective, possible relationships between the CAF parameters (dependent variables) and selected subject-specific parameters (predictor variables) were sought. A priori, all the possible latter parameters that might affect the CAF were identified by logical reasoning and with reference to the literature [9,10]. These parameters were: body mass (M), stature (H), body mass index ($BMI = M/H^2$ [18]), estimate of body surface (obtained as $S = 7.184 \times M^{0.425} \times H^{0.725}$ [19], measured in SI units) which is linearly related to the mean cardiac output [19], mean heart frequency (f), and the product of body surface with the mean heart frequency ($S \times f$). The latter parameter was chosen as related with the force applied to the blood mass by the heart. In fact, this force is proportional to the product of the mass in motion times its acceleration. The former variable is given by the cardiac output, proportional to body surface, divided by the cardiac frequency (S/f), while the acceleration may be assumed to be related with the square of this frequency (f^2).

The inspection of the relevant scatter plots indicated either relatively linear patterns or a near-zero correlation. Thus, the linear regression coefficients ($\hat{a}$ = intercept and $\hat{b}$ = slope) were estimated. The linear correlation between the variables involved was measured using the Pearson product–moment correlation coefficient (r). The linear correlation was deemed reliable if this coefficient, subjected to a two-tailed significance test with 18 (sample size -2) degrees of freedom and $P \leq 0.05$, was greater or equal to 0.444. The average prediction error for the regression equations was assessed using the standard error of the estimate (SEE).

Sex was not taken into consideration since there were no clues that it would affect the quantities under analysis.

3. Results

The mean cardiac frequency of the 20 subjects was found in the range 1.04–1.96 beats per second, with an intra-subject CV in the range 1–4%.

Typical averaged CAF component time histories are depicted in Fig. 1 together with the relevant 95% confidence limits. The related resultant vector amplitude is given in Fig. 2. With the exception of F_z, the overall pattern of these time functions was relatively repeatable across all examined subjects, although peak amplitudes and the relevant time of occurrence underwent variations, as will be highlighted later. As far as F_z was concerned, 12 of the 20 subjects showed a pattern similar to the one shown in Fig. 1. The other subjects, although characterised by similar peak-to-peak values, had patterns that could not be classified.

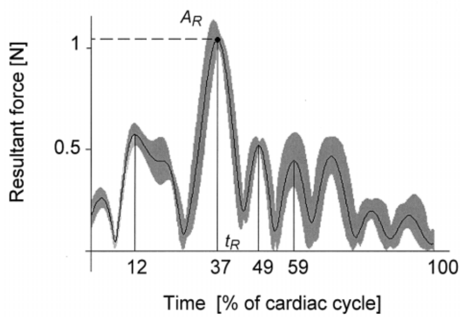

Fig. 2. CAF resultant amplitude with 95% confidence limits, during a cardiac cycle, as obtained in the same subject as in Fig. 1. The peak, with an amplitude and occurrence time included in the CAF parameters (see Table 1), is also indicated.

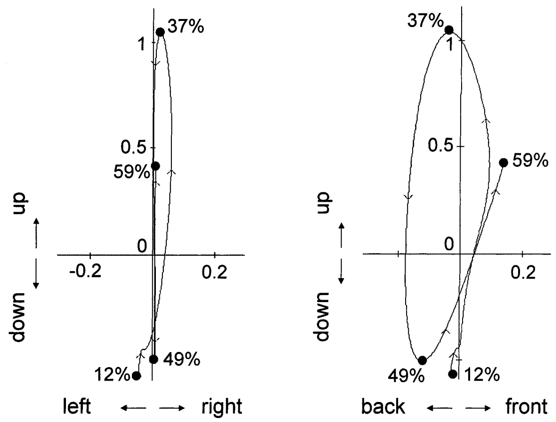

Fig. 3. Cardiac activity resultant force as projected onto the sagittal and frontal planes in the time interval ranging from 12 to 49% of the cardiac cycle. Filled circles indicate the time instants when the resultant force peaks indicated in Fig. 2 occur. These data were obtained from the same subject as in Figs. 1 and 2.

An example of the way the orientation in the sagittal and frontal planes (*xy* and *zy* planes, respectively) of the CAF vector change over time, in the time interval during which the relevant major peaks occur (see Fig. 2), is shown in Fig. 3. The time instants when the peaks indicated in Fig. 2 occurred are also indicated in Fig. 3. These patterns were similar in all subjects. The plots of the projections onto the *zx* (transverse) plane did not show a consistent pattern and carried no reliable information.

The most repeatable events across the subjects were identified by inspecting the CAF component and resultant time histories. The antero-posterior component was characterised by a repeatable triphasic wave preceded and followed by relatively small oscillations that occupied approximately the first 15% and last 20% of the cardiac cycle, respectively, and that were not consistent among subjects. The vertical component showed the most repeatable pattern with special reference to the triphasic wave located approximately between 10 and 65% of the cardiac cycle. The already mentioned 12 subjects who had a consistent mediolateral CAF component showed a repeatable biphasic wave located between 5 and 40% of the cardiac cycle. Consistently with the fact that, on average, F_y had peak values that were 4 times and 16 times larger than F_x and F_z, the resultant CAF amplitude plot displayed the same characteristics of repeatability as the former component.

The peaks of the above-illustrated repeatable events are indicated and labelled in Figs. 1 and 2. Peak amplitudes (A_{ax}, A_{bx}, A_{cx}, A_{ay}, A_{by}, A_{cy}, A_{az}, A_{bz}, A_R), taken as absolute values, and the related occurrence times (t_{ax}, t_{bx}, t_{cx}, t_{ay}, t_{by}, t_{cy}, t_{az}, t_{bz}, t_R), were included in the CAF parameters together with the angles ($\varphi_{xy}(t_R)$, $\varphi_{yz}(t_R)$, $\varphi_{zx}(t_R)$), which described the orientation of the resultant CAF when it assumed its maximum value (A_R), and the rectified force impulses (I_x, I_y, I_z). The resultant CAF amplitude was simply described by its maximal value and occurrence time because the overall pattern was already represented in F_y. With reference to the F_z component, the above parameters were, of course, identified in the 12 subjects who had consistent patterns.

For all parameters, with the exception of $\varphi_{zx}(t_R)$ (ICC = 0.43), a distinctly better intra- than inter-individual repeatability was found (ICC values ranged from 0.75 to 0.98). The within-subject descriptive statistics, carried out on the CAF parameters over the six trials for each subject, gave CV values lower than 25% for the peak amplitudes, lower than 8% for the resultant CAF angles $\varphi_{xy}(t_R)$ and $\varphi_{yz}(t_R)$, and lower than 10% for the peak occurrence times. With the above-illustrated limitations relative to F_z and to φ_{zx}, these results confirmed that subject specific CAF patterns were identified.

The between-subjects descriptive statistics of the CAF parameters are given in Table 1. As can be seen, several parameters underwent large relative variations (CV) that could be associated either with a low signal-to-noise ratio or with their dependence on other subject-specific parameters. The latter circumstance was checked by linear correlation analysis and these results are also given in Table 1. The peak amplitudes and rectified impulse associated with the F_z component and A_{ax} showed no correlation with the selected predictor variable; in fact they had small values and presumably unfavourable signal-to-noise ratios. The same applied to the F_z peak occurrence times. The parameters $\varphi_{xy}(t_R)$, $\varphi_{yz}(t_R)$ and $\varphi_{zx}(t_R)$ showed no correlation. Parameter $\varphi_{zx}(t_R)$ had a large CV value which, as expected, indicates a low signal-to-noise ratio. The mean and S.D. of the values that the angles φ_{xy} and φ_{yz} assumed when the resultant CAF reached the other peaks indicated in Fig. 2 were also estimated.

4. Discussion

The results of this study provided information on the amplitude and direction of the internal reaction forces elicited by hemodynamics during a cardiac cycle that, with reference to a population of young able-bodied subjects, could be generalised as follows.

The antero-posterior CAF component is characterised by a triphasic major wave (Fig. 1) the peaks of which had values up to 0.40 N and showed a remarkable inter-subject variability (CV up to 50%); however, their timing was rather consistent (CV within 17%, Table 1). No correlation was found for the amplitude of the first peak (A_{ax}), mostly acting frontward, and this was very likely due to a low signal-to-noise ratio. The absolute values of the amplitudes of the second peak, acting backwards, and of the third peak, acting frontward, positively correlated with the BMI, although with weak reliability. The relevant regression equations allowed for the prediction of these variables with a SEE that was only 15% lower than the relevant value of S.D., i.e. with a 15% better reliability than the one obtained if the relevant mean value (m) were used. The latter circumstance confirmed the weakness of the correlation. The occurrence times of these three peaks, expressed as a percentage of T, reliably correlated positively with the mean cardiac frequency f and, using the regression equations, could be estimated with a SEE that was 50% better than S.D. The same temporal parameters, expressed in seconds, showed a consistent but moderate (i.e. small slope) negative correlation with respect to f. This means that the peak delays relative to the R wave decrease as the duration of the cardiac cycle decreases, but to a lesser percentage. This is consistent with a similar relationship found between cardiac cycle duration and systole duration [20].

The vertical component of the CAF showed a repeatable triphasic wave with peak-to-peak values in the range 1.3–3.0 N. The middle peak of this wave acted upwards and was, on average, larger than the others, which acted downwards, and had values in the range 0.6–1.7 N. Amplitudes (CV from 28 to 43%), were more dispersed than timing (CV within 18%). All amplitudes positively correlated with the predictor variable Sf and could be estimated with a SEE lower than 0.24 N (23% lower than S.D.). Relevant percentage occurrence times positively correlated with f and could be estimated with a SEE approximately 40% lower than S.D. As for the antero-posterior CAF component, absolute times moderately correlated negatively with f. The pattern found for the vertical CAF component very closely resembles that of the ballistocardiogram recorded in the cranio-caudal direction while subjects were supine. In particular, the timing of the triphasic

Table 1
Mean values (m), S.D. and CV of the CAF parameters for the 20-subject sample (12-subject sample for the CAF medio-lateral (z) component — see text for explanation)[a]

CAF parameters	m	S.D.	CV (%)	Predictor	$\hat{a}$	$\hat{b}$	r	SEE
A_{ax}	0.12	0.06	50	–	–	–	–	–
A_{bx}	0.20	0.07	35	BMI	−0.132	0.015	0.559	0.06
A_{cx}	0.16	0.07	46	BMI	−0.214	0.017	0.608	0.06
A_{ay}	0.70	0.30	43	Sf	−0.147	0.333	0.602	0.23
A_{by}	1.10	0.30	28	Sf	0.245	0.335	0.605	0.23
A_{cy}	0.53	0.20	37	Sf	0.095	0.171	0.475	0.17
A_{az}	0.03	0.01	23	–	–	–	–	–
A_{bz}	0.04	0.01	15	–	–	–	–	–
A_R	1.10	0.30	27	Sf	0.273	0.328	0.607	0.24
t_{ax}	27.0	4.5	17	f	10.41	11.275	0.659	3.2
t_{bx}	41.0	6.0	15	f	11.88	19.603	0.853	2.9
t_{cx}	54.8	8.9	16	f	11.02	30.020	0.659	3.9
t_{ay}	20.0	3.6	18	f	9.50	7.102	0.543	3.0
t_{by}	33.0	5.3	16	f	10.13	15.548	0.801	3.1
t_{cy}	46.0	7.8	17	f	9.23	25.215	0.860	4.0
t_{az}	12.8	3.6	28	–	–	–	–	–
t_{bz}	31.9	6.7	21	–	–	–	–	–
t_R	32.0	5.5	17	f	10.13	15.597	0.633	3.1
I_x	106.0	29.2	28	BMI	−108.29	9.742	0.880	13.5
I_y	406.0	99.7	25	S	−306.16	410.246	0.624	75.9
I_z	22.0	8.2	37	–	–	–	–	–
$\varphi_{xy}(t_R)$	89.2	4.4	5	–	–	–	–	–
$\varphi_{yz}(t_R)$	88.7	1.0	1	–	–	–	–	–
$\varphi_{zx}(t_R)$	96.5	24.8	26	–	–	–	–	–

[a] Relevant linear regression coefficients ($\hat{a}$ = intercept and $\hat{b}$ = slope) and predictor variable, the Pearson product–moment correlation coefficients (r), and the SEE are also reported when the correlation was reliable ($P \leq 0.05$, $r \geq 0.444$).

wave indicated above is in line with that of the ballisto-cardiographic IJK wave reported in the literature [21,22].

The medio-lateral component in 12 of the 20 subjects showed two major peaks, acting first to the left and then to the right. The relevant peak-to-peak values (in the range 0.05–0.10 N) were the smallest among the three CAF components and, consequently, gave rise to the lowest signal-to-noise ratio. The latter circumstance may have caused the fact that no correlation was found with any of the predictor variables.

The resultant CAF vector had an amplitude that underwent several oscillations during the cardiac cycle and reached its maximal value in the range 0.6–1.7 N. This value was positively correlated with *Sf* (SEE 20% lower than S.D.), as was the case with the selected peak amplitudes of F_y. The relevant percentage occurrence time fell between 25 and 40% of T. It was linearly correlated with f and the relevant regression equation estimated it with a SEE equal to 55% of S.D. With reference to the orientation in the sagittal and frontal planes of the CAF vector when it assumed the peak values indicated in Fig. 3, the following was noticed. The first peak consistently acted downward ($\varphi_{xy} = -91.2$ (3.9), $\varphi_{yz} = -90.7$ (1.5)). The second and largest peak upward and, in most cases, rightward (Table 1). The third peak acted downward ($\varphi_{xy} = -98.8$ (18.6), $\varphi_{yz} = -89.9$ (4.8)). The fourth peak acted upward and frontward although the latter direction displayed great variability ($\varphi_{xy} = 47.2$ (21.9), $\varphi_{yz} = 56.2$ (54.2)). Directions that are not mentioned were found to be inconsistent.

As seen above, the peak amplitudes of the antero-posterior CAF component correlated with the BMI, while the vertical component and the resultant amplitudes correlated with the product of body surface with the mean heart frequency ($S \times f$). We do not have an explanation for this difference. However, the correlation of the resultant force, which is mostly vertical, with respect to the predictor variable *Sf* may be justified by the fact that this variable is proportional to the force applied to the blood mass by the heart, and vice versa, and blood is ejected upwards.

As indicated by the parameters I_x, I_y, and I_z, throughout the cardiac cycle the CAF impulse acted almost totally in the sagittal plane (Table 1). The rectified impulse along the vertical axis positively correlated with S and, in turn, with the cardiac output. The same correlation was found by Stürm et al. [10]. However, these authors could not identify the other CAF components and, therefore, their possible correlations. The antero-posterior impulse correlated with the BMI consistently with the relevant peak amplitudes. It also correlated with S, although with a lower correlation coefficient ($r = 0.576$).

A final remark concerns a preliminary contribution to answering the third question posed in the introduction relating to the possibility of identifying the effect of blood dynamics on the COP trajectories. In fact, restricting the analysis to the sagittal plane, to a quasi-static approach and representing the human body as an inverted pendulum with an elastic hinge at the ankle, it is easy to calculate the torsional elastic stiffness of this hinge while the subject is standing in equilibrium. Assuming that the line of gravity falls a few centimetres in front of the hinge and that a reasonable body weight for an adult subject is considered, then the elastic coefficient would be in the range $0.5 \times 10^3 – 10^3$ Nm/rad. These values are roughly in agreement with those found by other authors [23,24]. The results provided by this study indicate that the CAF moment with respect to the ankle may reach the value of 0.4 Nm. This moment, under the above-mentioned circumstances, would entail a displacement of the COM and of the COP, within the duration of a cardiac cycle, approximately between 0.5 and 1 mm. Note that under dynamic circumstances, the latter figures may result in an overestimate. Whether displacements of this order of magnitude are relevant or not in the present context is an issue for further study.

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