GLOBAL BIOMECHANICAL EVALUATION DURING WORK AND DAILY-LIFE ACTIVITIES

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Abstract: Advances in technology in the last decades have provided the opportunity to observe human behaviour in the three dimensional space with great spatial accuracy. Optoelectronic techniques for measurements of human motions have been developed. However, it is found that, in the work environments, these methods are complicated to set up and can only easily be applied in laboratory. On the other hand, electronic sensors such as accelerometers and gyroscopes, have been developed and applied to solve the relevant outdoor application problems of the image-based methods. These sensors have been evaluated for the 3D measurement of trunk, lower and upper segments, during posture, walking and rising from a chair, in both normal and pathological conditions. In the present study we used a device including accelerometers and gyroscopes in order to calculate the angular behaviour of the pelvis on the sagittal, frontal and horizontal plane, during the following tasks: walking, gait initiation, gait termination, seat-to-stand and stand-to-seat, squat, standing anterior and lateral reaching and grasping, anterior and lateral trunk flexion and trunk rotation. The assessment of pelvis during posture and movement is important in improving our understanding of the motor strategies at work and preventing injuries (i.e. low back pain) and mechanical whole body fatigue. The calculated angles were compared to that computed by a high-quality optical motion analysis system (SMART-E System, BTS, Milan, Italy) consisting of eight infra-red cameras (operating at 120 fps) to detect the movements in three-dimensional space of three retro-reflective markers (15 mm diameter). For the comparison of the Range of Motions (ROMs) we used the root mean squared error (RMS) whereas the Coefficient of Multiple Correlation (CMC) was used to evaluate overall waveform similarity of instantaneous angle curves. Preliminary results showed a high similarity between the extracted angle tracks (anterior-posterior behaviour on the sagittal plane, pelvic obliquity and intra-extra rotation of the pelvis) in all of the acquired tasks. We also found low errors in the computation of the corresponding ROMs. This study suggests to apply an accurate, inexpensive and simple method to measure the kinematics of the pelvis during common work and daily-life activities.

1 INTRODUCTION

Advances in technology in the last decades have provided the opportunity to observe human behaviour in the three dimensional space with great spatial accuracy. Image-based methods for the measurement of human motion have been developed, such as optoelectronic techniques (Medved, 2001; Cappozzo, 2005). However, it is

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found that, in work environments, these methods are complicated to set up and can only easily be applied in laboratory. On the other hand, electronic sensors able to provide orientation based on accelerometers and gyroscopes have been developed and applied to solve the relevant outdoor application problems of the image-based methods. These sensors have been used and evaluated for the 3D measurement of trunk, lower and upper segments, during posture, walking and rising from a chair, in both normal and pathological conditions (Pfau, 2005, Lau, 2008, Plamondon, 2007, Coley, 2007, Veltink, 2007, Boonstra, 2006, Zijlstra, 2008).

In the present study we compared the angular behaviour of the pelvis in the sagittal, frontal and horizontal plane calculated with a wearable inertial device including triaxial accelerometers and triaxial gyroscopes with that computed with a high precision and accuracy optoelectronic motion analysis system. We recorded the angle trajectories and excursions during the following tasks: standing anterior reaching and grasping, standing oblique reaching and grasping, standing oblique opposite reaching and grasping, anterior trunk flexion and sit to stand.

The assessment of pelvis during posture and movement is important in improving our understanding of the motor strategies at work and preventing injuries (i.e. low back pain) and mechanical whole body fatigue.

2 MATERIALS AND METHODS

Ten healthy male subjects (mean age 38 ± 4 years, range 20-55 years) were enrolled. All gave their written informed consent after receiving a full explanation of the study, which conformed to the requirements of the Declaration of Helsinki.

We used a Wi-Fi transmission miniaturized device integrating an accelerometer and a gyroscope (MicroStrain 3DM-GX2, MicroStrain, Inc., Williston, USA) placed directly on the skin over the sacrum. The device offers a range of output data quantities from fully calibrated inertial measurements to computed orientation estimates. All quantities are fully temperature compensated and corrected for sensor misalignment. The angular rate quantities are further corrected for G-sensitivity and scale factor non-linearity to third order.

The extracted curves and the calculated angles were compared to those simultaneously acquired by a high-quality optical motion analysis system (SMART-E System, BTS, Milan, Italy, Ferrigno and Pedotti 1985) consisting of eight infra-red ray cameras (operating at 120 fps) to detect the movements in three-dimensional space of three retro-reflective markers placed on the skin over the sacrum and the right and left anterior superior iliac spinae. Data processing was performed using Analyzer software (BTS, Milan, Italy).

Before starting formal measurements, all subjects did a practice session to familiarize themselves with the experimental procedure and with the tasks consisting of eleven movements performed in a quiet room with normal indoor temperature and lighting. The performed tasks were: standing anterior reaching and grasping, standing oblique reaching and grasping, standing oblique opposite reaching and grasping, anterior trunk flexion and sit to stand. Standing reaching and grasping tasks have been performed with the subject starting from a standing posture, with the trunk kept upright, left and right arm lying alongside the body and performing the movement, in a natural fashion. In the anterior reaching and grasping the subjects picked up, with the right hand, a cylinder (diameter, 3 cm; height, 6 cm; weight, 300 g) positioned on a shelf in line (on the anterior direction) and at the same height of the right shoulder, and returned the cylinder to the starting position. The oblique and the oblique opposite reaching and grasping tasks were performed in the same manner of the anterior reaching and grasping but with the object positioned at ±45° with respect to the anterior direction. The anterior trunk flexion was performed through a maximal anterior flexion of the trunk. In the sit to stand movement the subjects were seated comfortably on a chair and got stand up in a natural manner and at their preferred velocity. Ten cycles were recorded for task and each person. Angular excursion data were normalized to the movement duration and reduced to 100 samples. For the comparison of the Range of Motions (ROMs) we used the root mean squared error (RMS):

$$RMS = \sqrt{\frac{\sum_{1=1}^{N}\left(x_{o,i} - x_{d,i}\right)^2}{n}} \tag{1}$$

whereas the Coefficient of Multiple Correlation (CMC), i.e. the positive square root of the adjusted coefficient of multiple determination (Kabada et al. 1989, Steinwender et al. 2000) by means of the following formula:

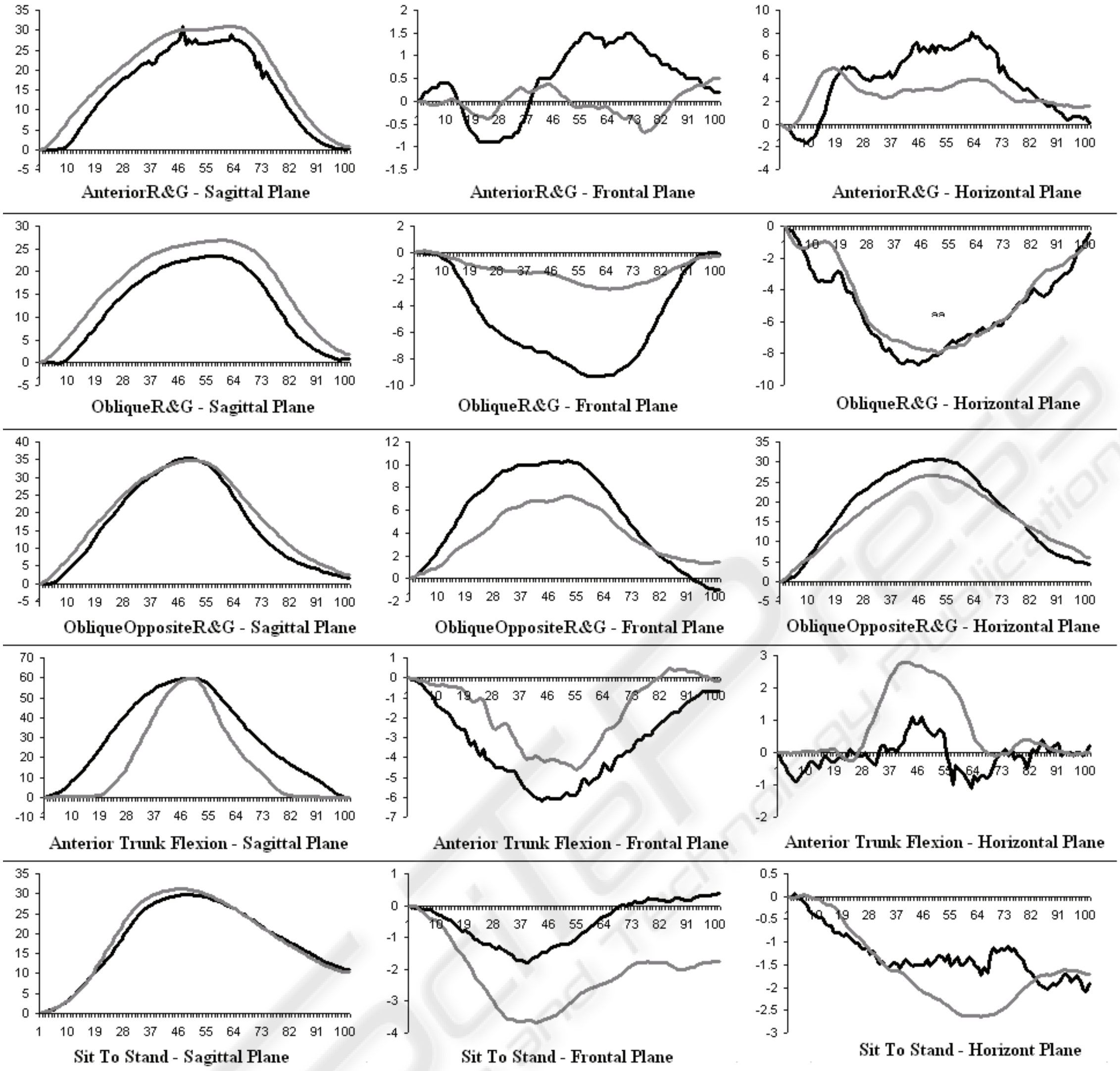

Figure 1: First column: pelvic angular behaviour in the sagittal plane (pelvic tilt); second column: pelvic angular behaviour in the frontal plane (pelvic obliquity); third column: pelvic angular behaviour in the horizontal plane (pelvic intra-extra rotation). 1th, 2th, 3th, 4th and 5th rows show the pelvic mean angular curves during the standing anterior reaching and grasping, the standing oblique reaching and grasping, the oblique opposite reaching and grasping, the anterior trunk flexion and the sit to stand respectively. In black and grey, curves acquired and computed by the high-quality optical motion analysis system and the Wi-Fi transmission miniaturized device respectively. On x-axis and y-axis are reported percentage cycle time duration and degrees respectively.

Table 1: ROMs (mean±standard deviation) calculated by the optoelectronic motion analysis system (Opt), by the wearable inertial device (G+A), the paired *t* test P value, the RMS and the CMC for each motor task and for sagittal, frontal and horizontal plane.

	Sagittal					Frontal					Horizontal				
	Opt	G+A	P	RMS	CMC	Opt	G+A	P	RMS	CMC	Opt	G+A	P	RMS	CMC
AnteriorR&G	27.7±5.7	31.7±8.1	0.5	4.0	0.96	0.3±0.6	1.7±0.3	0.023	1.4	0.50	8.2±1.7	6.6±0.8	0.2	2.0	0.61
ObliqueR&G	23.5±1.5	26.9±1.8	0.066	3.4	0.94	6.3±4.6	3.3±1.2	0.336	5.2	0.82	9.9±2.0	9.7±1.5	0.897	1.6	0.96
ObliqueOppositeR&G	36.2±3.2	36.0±3.1	0.942	5.3	0.98	10.7±1.0	7.7±1.2	0.029	3.5	0.84	31.5±0.7	28.0±0.9	0.006	3.5	0.96
Anterior Trunk Flexion	59.9±0.6	60.4±2.1	0.712	1.6	0.88	6.3±0.2	5.3±1.7	0.369	1.8	0.77	1.9±0.3	3.4±0.4	0.007	1.5	0.31
Seat To Stand	30.6±2.0	31.6±1.6	0.536	3.4	0.99	1.9±0.4	4.0±1.0	0.013	2.1	0.70	-1.2±0.9	3.0±0.2	0.001	4.1	0.76

$$CMC = \sqrt{1 - \frac{\frac{1}{T(N-1)}\sum_{i=1}^{N}\sum_{t=1}^{T}\left(y_{it} - \bar{y}_t\right)^2}{\frac{1}{TN-1}\sum_{i=1}^{N}\sum_{t=1}^{T}\left(y_{it} - \bar{y}\right)^2}} \qquad (2)$$

where T=100 (number of time points within the cycle), N=2 (number of curves), y_{it} is the value at the tth time point in the ith cycle, $\bar{y}_t$ is the average at time point t over N cycles:

$$\bar{y}_t = \frac{1}{N}\sum_{i=1}^{N} y_{it} \qquad (3)$$

and $\bar{y}$ is the grand mean of all y_{it}:

$$\bar{y} = \frac{1}{NT}\sum_{i=1}^{N}\sum_{t=1}^{T} y_{it} \qquad (4)$$

CMC was used to evaluate overall waveform similarity of instantaneous angle curves: the closer to 1 the CMC, the more similar the waveforms. The statistical analysis was performed using SAS 8.2 (SAS Institute Inc., Cary, NC, USA). A paired *t* test was applied in order to compare ROMs calculated by the two techniques. *P*-values less than 0.01 were considered statistically significant.

3 RESULTS

Results are summarized in Figure 1 and in Table 1. 1th, 2th, 3th, 4th and 5th rows of Figure 1 show the pelvic mean angular curves during the standing anterior reaching and grasping, the standing oblique reaching and grasping, the oblique opposite reaching and grasping, the anterior trunk flexion and the sit to stand respectively. The 1th, 2th and 3th column of Figure 1 show the pelvic angular behaviour in the sagittal plane (pelvic tilt), in the frontal plane (pelvic obliquity) and in the horizontal plane (pelvic intra-extra rotation). In black and grey, curves acquired and computed by the high-quality optical motion analysis system and the Wi-Fi transmission miniaturized device respectively. Table 1 shows the ROMs (mean±standard deviation) calculated by the optoelectronic motion analysis system (Opt), by the wearable inertial device (G+A), the paired *t* test P value, the RMS and the CMC for each motor task and for sagittal, frontal and horizontal plane.
The results showed high similarity between the extracted angle curves (Figure 1, Table 1) with respect to the pelvic tilt on the sagittal plane (CMC>0.88), the pelvic obliquity on the frontal plane (CMC>0.70 except for the anterior reaching and grasping) and the pelvic intra-extra rotation on the horizontal plane (CMC>0.61 except for the anterior trunk flexion). We also found low root mean square errors in the computation of the corresponding ROMs in the sagittal plane (RMS≤5.3). Statistically significant differences in the calculated ROMs were found only for the standing oblique opposite reaching and grasping, anterior trunk flexion and sit to stand on the horizontal plane (P>0.01).

4 CONCLUSIONS

We compared data acquired and computed by two different complementary technologies: a wearable inertial device and an optoelectronic system. The former allows a simple setup and outdoor acquisitions (e.g. work environment); the latter represent the kinematic gold standard acquisition system but it is not simple to set up in work environment. The use of wearable inertial devices can be considered very useful when a simple biomechanical human global approach is needed (e.g. study of the human mechanical energy expenditure, of the whole-body stiffness and of the centre of mass behaviour). Our results suggest also the use of these devices in work environment applications, where specific segmental analyses are needed, such as the study of the pelvic behaviour. In these conditions, they yield good precision and accuracy values on those measures of angular components that present high magnitude of ROMs, such as sagittal and frontal component of the our study. Planes on which ROMs have low amplitudes don't show good similarity of curves and present high root mean square errors.

Furthermore, this study suggests applying these accurate, inexpensive and easy to use methods to measure the kinematics of the pelvis during common work and daily-life activities.
These devices could as well be used by a biofeedback approach, which is widely considered as a valid tool in various rehabilitation contexts (Nelson, 2007). Kinematic-based (Van Vliet, 2006) biofeedback frameworks have been proposed for a routine inclusion in rehabilitation protocols. On the other hand, there aren't evidences on the use of these devices in the return back to work environment of workers after injuries. As these techniques share the advantage of being suitable for workers' self-administration, they may also be suitable for use in

telerehabilitation, which is not yet widespread mainly due to the unavailability of specific devices, validated protocols and appropriate operators' educational programs. Indeed, the use of these methods at home or at work may allow workers after injuries to gain control over their own motor recovery, to increase frequency and duration of physical training and to improve personal involvement and satisfaction in the rehabilitation program. A miniaturized, wearable device for kinematic biofeedback, interfaced with a telerehabilitation platform, may improve the quality of rehabilitation due to a faster getting back of workers to their usual environments, with a beneficial effect on quality of life, a minimization of lost opportunity costs for employers, who can be treated onsite reducing absence at work, and a decrease of the economic burden for the healthcare system. These techniques will be relevant for the National Health Service in order to provide data about the feasibility of rehabilitation treatment transfer from hospitals to work settings. Such transfer may allow the National Health Service to reduce costs and ameliorate the managing of the resources employed in this context.

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