Brice, Sara M., Phillips, Elissa J., Millett, Emma L., Hunter, Adam, and Philippa, Bronson (2020) ***Comparing inertial measurement units and marker-based biomechanical models during dynamic rotation of the torso.*** **European Journal of Sport Science, 20 (6) pp. 767-775.**

https://doi.org/10.1080/17461391.2019.1666167

European Journal of Sport Science

ISSN: 1746-1391 (Print) 1536-7290 (Online) Journal homepage: https://www.tandfonline.com/loi/tejs20

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To cite this article: Sara M. Brice, Elissa J. Phillips, Emma L. Millett, Adam Hunter & Bronson Philippa (2019): Comparing inertial measurement units and marker-based biomechanical models during dynamic rotation of the torso., European Journal of Sport Science, DOI: 10.1080/17461391.2019.1666167

To link to this article: https://doi.org/10.1080/17461391.2019.1666167

Publisher: Taylor & Francis & European College of Sport Science

Journal: *European Journal of Sport Science*

DOI: 10.1080/17461391.2019.1666167

Title: Comparing inertial measurement units and marker-based biomechanical models during dynamic rotation of the torso.

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Abstract

Inertial measurement units (IMUs) enable human movements to be captured in the field and are being used increasingly in high performance sport. One key metric that can be derived from IMUs are relative angles of body segments which are important for monitoring form in many sports. The purpose of this study was to a) examine the validity of relative angles derived from IMUs placed on the torso and pelvis; and b) determine optimal positioning for torso mounted sensors such that the IMU relative angles match closely with gold standard torso-pelvis and thorax-pelvis relative angle data derived from an optoelectronic camera system. Seventeen adult participants undertook a variety of motion tasks. Four IMUs were positioned on the torso and one was positioned on the pelvis between the posterior superior iliac spines. Reflective markers were positioned around each IMU and over torso and pelvis landmarks. Results showed that the IMUs are valid with the root mean square errors expressed as a percentage of the angle range (RMSE%) ranging between 1% and 7%. Comparison between the IMU relative angles and the torso-pelvis and thorax-pelvis relative angles showed there were moderate to large differences with RMSE% values ranging between 4% and 57%. IMUs are highly accurate at measuring orientation data; however, further work is needed to optimize positioning and modelling approaches so IMU relative angles align more closely with relative angles derived using traditional motion capture methods.

Keywords: 3D analysis; Biomechanics; Engineering; Measurement.

Introduction

In elite sport, key metrics on athlete movement dynamics are widely used by coaches and sports scientists to examine an athlete's technique. This information can be used to improve performance and identify movement deficiencies that can lead to increased incidences of injury. Traditionally, movement dynamics have been captured using optoelectronic systems, in which multiple cameras are used to locate retro-reflective markers in three-dimensional space. However, these systems can only be used in constrained environments and support a limited capture volume. Consequently, certain activities are very difficult to record using optical techniques and there is demand for an alternative technology.

Inertial measurement units (IMUs) have recently been adopted to measure movement in both clinical and sporting settings. These systems do not require the activity to be confined to a specific capture volume. IMUs contain accelerometers, gyroscopes, and magnetometers and when placed on a body segment can be used to determine the segment's three-dimensional orientation. This is done by fusion of the sensor data using algorithms such as a Kalman or Madgwick filter (Kalman, 1960; Madgwick, Harrison, & Vaidyanathan, 2011). Furthermore, when IMUs are placed on adjacent body segments they can be used to determine the angles between segments using an Euler angle rotation sequence (Grood & Suntay, 1983). These angles represent joint angles and are commonly used to examine human movement. A number of previous studies have investigated the validity and accuracy of using IMUs to measure a number of different joint angles (Bauer et al., 2015; Blair, Duthie, Robertson, Hopkins, & Ball, 2018; Brice, Hurley, & Phillips, 2018; Cottam et al., 2018; Fantozzi et al., 2016; Walgaard, Faber, van Lummel, van Dieën, & Kingma, 2016; Wong & Wong, 2008). Results from these studies have been varied and are specific to the movements that were examined and the sensors that were employed. It is important that validation be undertaken for each different movement of interest, and for each type of sensor and sensor placement protocol (Cuesta-Vargas, Galán-Mercant, & Williams, 2010).

One movement that is of interest in a number of sporting disciplines that could be measured in-field with IMUs is relative movement of the torso (upper and lower) to the pelvis (torso-pelvis angles). This type of movement is pertinent within the athletic throwing disciplines, cricket fast bowling, handballing in Australian rules football, rowing, and golf. The torso-pelvis angles have been previously shown to be related to performance (Brice, Ness, Everingham, Rosemond, & Judge, 2018; Horan, Evans, Morris, & Kavanagh, 2010; Leigh, Gross, Li, & Yu, 2008; Leigh & Yu, 2007; Myers et al., 2008; Parrington, Ball, & MacMahon, 2014; Portus, Mason, Elliott, Pfitzner, & Done, 2004) and increased injury incidences (Elliott, 2000; Foster, John, Elliott, Ackland, & Fitch, 1989; Ng, Campbell, Burnett, Smith, & O'Sullivan, 2015; Wilson, Gissane, Gormley, & Simms, 2013). Typically when movement of these body segments is measured using an optoelectric system, a model is employed that requires a number of assumptions to be made and uses multiple marker locations and anatomical information. An example is the Plug-in Gait model (Oxford Metrics, Oxford, UK).

There has been little work undertaken to identify how closely IMU-measured relative angles align with the angles computed using marker locations and a model, here forth referred to as model angles. Previous work has focused on comparing IMU relative angles with relative angles computed from the optically measured orientations of the IMUs (determined by placing markers around each IMU). While this shows how accurately orientation can be measured using an IMU, it does not examine whether the IMU relative angles are comparable with model angles. To the authors' knowledge, only two previous studies have compared IMU torso-pelvis relative angles with those computed using a model and both found poor agreement (Brice et al., 2018; Cottam et al., 2018). However, in the aforementioned studies only one torso IMU location was examined in each study. It is possible different locations may result in relative angles that match model angles more closely.

The purpose of this present study was to examine the validity of relative angles computed using the IMeasureU BlueThunder V1.0 IMUs (Oxford Metrics, Oxford, UK) and examine whether there is an optimal position for the torso sensor that results in relative angles that closely match those computed using a traditional marker based model. In our previous study (Brice et al., 2018) that examined this type of movement using these sensors, only one torso sensor location was examined (spinous process of the third thoracic vertebra – T3) and only transverse plane relative rotation was examined. Conversely, this study aims to identify a location for the torso IMU that maximises the accuracy of relative angles for all three anatomical reference planes when compared to a traditional marker based model.

Methods

Participants and testing preparation

Seventeen healthy adults (8 male and 9 female) participated in this study which was given ethical approval by the James Cook University Human Research Ethics committee. Each participant gave written informed consent prior to data collection. An *a priori* sample size calculation (Erdfelder, Faul, & Buchner, 1996) was used to confirm that seventeen participants was sufficient. In this calculation, statistical power was set at 80% and a large effect was chosen ($r = 0.8$) based previously collected data (Bauer et al., 2015; Blair et al., 2018; Walgaard et al., 2016; Wong & Wong, 2008).

Five IMeasureU BlueThunder V1.0 IMUs (Oxford Metrics, Oxford, UK) were positioned on each participant. Each IMU consisted of tri-axial accelerometers (± 16 g), gyroscopes (± 2000 °/s), and magnetometers (± 1200 mT). Four IMUs were positioned on the torso and one on the pelvis between the posterior superior iliac spines (Figure 1). Three of the torso sensors were positioned on the posterior side, in line with the vertebral column at the following heights: immediately below the spinous process of the 7^{th} cervical vertebra (C7 sensor), over the spinous process of the 2^{nd} thoracic vertebra (T2 sensor), and in line with the most inferior points of the scapulae (approximately the spinous process of the 7^{th} thoracic vertebra – T7 sensor). The final torso sensor was positioned on the anterior side of the torso immediately below the sternum jugular notch (sternum sensor). Each of the torso mounted IMUs were affixed to inflexible plastic boards and were surrounded by three retro-reflective markers referred to as marker triads (Figure 1). The pelvis IMU was mounted directly on the participant's skin and surrounded by three skin mounted retro-reflective markers. The markers surrounding the IMUs were used to determine the gold standard orientation of each IMU. This was considered to be a gold standard as marker positions were measured using an optoelectronic camera system (Cuesta-Vargas at al., 2010) and placement of the markers on the inflexible plastic boards and rigid areas of the pelvis meant there was little to no skin movement artefact (Cappozzo, Cantani, Dalla Croce, & Leardini, 1995).

The most proximal markers of some IMU marker triads were positioned to align with anatomical landmarks. Specifically, the C7 proximal marker was positioned over the spinous process of C7 and the sternum proximal marker was positioned over the sternum jugular notch. Other retro-reflective markers were located at the following positions: xiphoid process, most inferior points of the scapulae, left and right anterior superior iliac spines (ASIS), and left and right posterior superior iliac spines (PSIS) which were two of the markers that made up the triad surrounding the pelvis sensor (Figure 1).

*** Figure 1 near here***

Data collection

Five trials were collected for each participant. Three trials focused on examining relative rotation of the torso to the pelvis in each anatomical reference plane where subjects were instructed to rotate at a self-selected slow speed (average movement speed was 146 ± 61 °/s). Two trials focused on examining the transverse plane rotation at a fast speed (average movement speed was 416 ± 21 °/s). Transverse plane rotation was investigated more closely given its strong association with both performance and injury in a number of sporting disciplines. All rotations were slower than the maximum gyroscope range of 2000 °/s. For the slow rotations, rotation within each plane was performed as a stand-alone trial where three full rotations within that plane were performed. A single rotation was defined as rotation to the left and right in the transverse plane, flexion and extension in the sagittal plane, and left and right lateral flexion in the frontal plane. For the fast transverse plane rotations three fast full rotations were performed in one trial and a second trial was undertaken where two fast full rotations were performed with the participant instructed to come to a complete stop between each of the two rotations. All rotation speeds and ranges were self-selected with the only instruction being that a faster speed was needed for the fast transverse plane rotations. Prior to and immediately after all trials, participants were instructed to stand stationary. This resulted in each trial having a period of no movement at the start and conclusion of data collection.

During the five trials, IMU data and marker positions were collected simultaneously. Each IMU logged data to an on-board SD card at 500Hz. The IMUs were operated and synchronised using the manufacturer's custom software applications and their recommended procedures. Marker positions were collected at 250Hz using a 20 camera Vicon Vantage camera system (Oxford Metrics, Oxford, UK). Vicon data were logged using Vicon Nexus v2.6 (Oxford Metrics, Oxford, UK). A seventh IMU sensor was used during data collection to enable the motion capture data to be synchronized with the IMU data (sync IMU). The sync IMU was not affixed to the participant but was instead placed directly against an electromagnet. At the start of each trial (prior to any movement) three magnetic pulses from the electromagnet were simultaneously measured via the analogue input of the Vicon system and the magnetometers of the sync IMU. Timing of these pulses were used to align data during post-processing.

Data processing

Marker trajectories were smoothed in Vicon Nexus v2.6 using a Woltring filter with a mean standard error of 9 mm, defined following a residual analysis (Winter, 2009). Trajectory data were then further processed in Matlab (Mathworks, Natic, USA) where custom scripts were used to model the thorax (or upper torso), torso, and pelvis using previously defined methods (Besier, Sturnieks, Alderson, & Lloyd, 2003; Campbell, Lloyd, Alderson, & Elliott, 2008). The pelvis was modelled using the ASIS and PSIS markers. The torso was modelled using the markers on the xiphoid process, both scapulae, and the mid-pelvis location where mid-pelvis was derived from the ASIS and PSIS marker locations. The thorax was modelled using the markers on the sternum jugular notch, C7, xiphoid process, and both scapulae. The origins of the pelvis, torso, and thorax were then used to compute relative angles.

The relative angle between the thorax and pelvis (referred to as model thorax-pelvis relative angle) and the relative angle between the torso and pelvis (referred to as model torso-pelvis relative angle) were computed and then exported as an Euler angle rotation sequence (Grood & Suntay, 1983). The International Society of Biomechanics proposed sequence was used where the order was ZXY (Wu et al., 2002; Wu et al., 2005). Marker data were also used to compute the gold standard orientation of each IMU to enable a direct comparison with the orientations calculated from each IMU. These orientations were also used to compute the relative angle between each of the four torso mounted IMUs and the pelvis IMU using the same Euler rotation sequence. We call these measurements the gold standard relative angle data.

The raw IMU sensor data were post-processed in Matlab (Mathworks, Natic, USA) using custom scripts. Firstly, the magnetometer data were calibrated using an ellipsoid fitting procedure (Kok & Schön, 2016). A Kalman filter (Kalman, 1960; Maximov, 2018) was used to calculate the orientation of each IMU. The Kalman filter automatically estimates and corrects for any gyroscope bias. The calculated orientations were then transformed from their initial reference frame (which is based upon the gravity vector and the Earth's magnetic field) to the laboratory reference frame. We calculated the difference between the two reference frames at the start of the trial. Specifically, the lab frame quaternions were calculated as,

$$q_{IMU}^{Lab}(t) = q_{Vicon}^{Lab}(0) \left(q_{IMU}^{Earth}(0)\right)^{*} q_{IMU}^{Earth}(t)$$

where superscripts indicate the reference frame, subscripts indicate the source of the measurement, and the parameter t represents time with $t = 0\ s$ being the start of the trial. These data were used to compute the relative angle between the torso and pelvis using the same Euler angle rotation sequence used with the marker data. There were four torso mounted IMUs meaning there were four different sets of relative angle data computed.

In summary, three sets of relative angles were calculated:

a) Torso IMUs relative to the pelvis IMU according to the Kalman filter;
b) Torso IMUs relative to the pelvis IMU according to the optoelectronic system; and
c) Model thorax and torso relative to the model pelvis according to the optoelectronic system.

The comparison of (a) and (b) was done to assess the accuracy of the IMU sensors. Comparison of (a) and (c) was done to assess which torso position resulted in data that most closely matches the angles calculated using traditional modelling methods.

Statistical analyses

For each trial, the coefficient of determination (r^2) and root mean square error (RMSE) were determined to examine the precision and accuracy of the IMUs (Bauer et al., 2015). RMSE as a percentage of the angle range (RMSE%) was also computed. Only the primary movement directions were examined within each trial. Specifically, for the sagittal plane rotations only the sagittal plane angles were examined, for the frontal plane rotations only the frontal plane angles were examined, and for the transverse plane rotations only the transverse angles were examined. Bland-Altman biases and upper and lower 95% limits of agreement were also computed (Bland & Altman, 1986) with a positive bias indicating there was an overestimation and a negative bias indicating there was an underestimation. The limits of agreement were equal to 1.96 times the standard deviation of the differences between the IMU and gold standard above (upper limit) and below (lower limit) the bias (Bland & Altman, 1986). Limits of agreement that are close to the bias indicate the IMU values closely agree with the gold standard values.

Results

Comparison of the relative angles measured using the IMUs and the marker triads surrounding the IMUs showed there was good agreement in all three anatomical reference planes for all trials (Tables 1 and 2). The root mean square errors expressed as a percentage of the angle range (RMSE%) ranged between approximately 1% and 7%. The smallest RMSE% values were observed for the frontal plane rotations followed by the transverse plane and then the sagittal plane. Bland-Altman biases indicate that the IMU relative angles were underestimated for the sagittal and frontal plane rotations as evidenced by the negative biases (Table 1). For both the slow and fast transverse plane rotation the Bland-Altman biases indicated that the IMU relative angles were overestimated as evidenced by the positive biases (Tables 1 and 2). The aforementioned RMSE% values and biases indicated the IMUs are valid for measuring relative angles In addition to the RMSE% and biases the coefficients of determination (r^2), which ranged between 0.97 and 1.00, indicating the waveforms of the IMU relative angles were closely matched with the gold standard relative angles.

There were moderate to large differences between the IMU relative angles and the model thorax-pelvis and model torso-pelvis separation angles in all three anatomical reference planes (Tables 1 and 2). RMSE% values ranged between approximately 4% and 57%. The largest differences were observed in the slow sagittal plane rotations where the average RMSE% was approximately 34% when compared with the model thorax-pelvis angle and approximately 49% when compared with the model torso-pelvis angle (Table 1). For the slow transverse plane rotations the average RMSE% was approximately 7% when compared with both model angles (Table 1) and approximately 5% for the fast rotations (Table 2). In the frontal plane the average RMSE% was approximately 4% when compared with the model thorax-pelvis angle and approximately 25% when compared with the model torso-pelvis angle (Table 1). While there are large differences in the magnitudes of the angles, particularly in the sagittal plane, there was close agreement in the waveforms with r^2 values ranging between 0.9 and 1.00 for the thorax-pelvis relative angles and 0.89 and 0.99 for the torso-pelvis relative angles (Tables 1 and 2; Figure 2).

When examining the RMSE% values (Table 1 and 2), there was no clear evidence to suggest that one sensor location was optimal for measuring angles in all three anatomical reference planes that were comparable with the model relative angles. For the transverse plane rotations (both fast and slow rotation trials) the sternum sensor produced angles that were most comparable with the model thorax-pelvis angles while the T2 sensor angles were most comparable with the model torso-pelvis angles (Tables 1 and 2). For the frontal plane rotations the T2 sensor angles were most comparable with the model thorax-pelvis angles and T7 sensor angles were most comparable with the model torso-pelvis angles (Table 1). For the sagittal plane rotations the T7 sensor angles were most comparable with both model relative angles (Table 1).

When compared with the model angles, the Bland-Altman biases indicated that for the transverse plane rotations the C7 and T2 sensors overestimated both of the model relative angles (Tables 1 and 2) while the T7 and sternum sensors typically underestimated both of the model relative angles (Tables 1). The reverse was true for the sagittal plane rotations where the C7 and T2 sensors overestimated both model angles and the T7 and sternum sensors underestimated both model angles. In the frontal plane the C7 and T2 sensors overestimated both of the model relative angles while the T7 and sternum sensors underestimated the model thorax-pelvis angle and overestimated the model torso-pelvis angles.

Discussion

The aim of this study was to assess the validity of the relative angles measured using IMUs and to identify if an optimal torso IMU position exists that results in angles that are comparable to those determined using a traditional marker based model. To assess the validity, relative angles computed using the IMUs were compared with gold standard relative angles determined by using triads of markers that surrounded each sensor. To assess for optimal IMU positioning, IMUs were placed at C7, T2, T7, and on the sternum and the angle of each sensor relative to the pelvis was compared with model relative angles.

The relative angles computed using the sensors were found to be highly valid as evidenced by the high level of agreement between the IMU relative angles and the relative angles computed using the marker triads. Similar levels of agreement have been observed by Bauer and colleagues (2015), who calculated relative angles using sensors located at L1 and T1 relative to a pelvis mounted sensor. For both sites they found the accuracy was higher in the frontal plane than in the sagittal which is in agreement with what was observed here (Table 1) and in the findings of Wong and Wong (2008). Bauer and colleagues (2015) suggested the error may have been due to a number of factors including skin artefact and camera noise which may have attributed to the error observed in this present study as well. Validity within the transverse plane was also high in this present study and the observed RMSE values (Tables 1 and 2) were comparable with existing literature (Walgaard et al., 2016).

An acceptable amount of error is an RMSE% is 10% (Walgaard et al., 2016). Nearly all RMSE% values in this present study fell below this threshold. RMSE% values greater than 10% were observed when IMU relative angles were compared with model relative angles.

Specifically, this was observed for the thorax-pelvis and torso-pelvis relative angles in the sagittal plane and the torso-pelvis relative angles in the frontal plane (Table 1). This indicates that while the sensors are highly valid, in some cases the relative angles they compute do not match with those that are output from marker based models. This agrees with the findings of Cottam and colleagues (2018) who observed significant differences between IMU and model relative angles. They suggested the differences could be due to assumptions that are made as part of the modelling procedures, such as body segments being rigid, or issues surrounding how capable IMUs are at representing the motion of a body segment. The findings of this present study supports the suggestion put forth by Cottam and colleagues (2018) that differences are due to modelling assumptions.

The findings of this present study give some insight into sensor positions that are better at estimating relative angles comparable to model angles for different types of movement. In summary, the following recommendations for IMU positions are suggested to improve the similarity between IMU relative angles and gold standard relative angles:

- For sports and activities where transverse plane rotation of the upper torso (thorax) relative to the pelvis is of interest: sensors positioned on the posterior pelvis and over the sternum,
- For sports and activities where transverse plane rotation of the lower torso relative to the pelvis is of interest: sensors positioned on the posterior pelvis and at T2,
- For sports and activities where measuring frontal plane rotation of the upper torso (thorax) relative to the pelvis is of interest: sensors positioned on the posterior pelvis and at T2.

In the cases of frontal and sagittal plane torso-pelvis rotation and sagittal plane thorax-pelvis rotation the acceptable RMSE% threshold of 10% was passed indicating high levels of disagreement between these IMU relative angles and the model relative rotations. Sensor locations outside the scope of this study may result in improved relative angles. Further research is suggested to develop a model that incorporates multiple torso sensors to calculate the relative angles in order to improve accuracy across all planes of movement.

While the findings of this study give insight into IMU use there were some limitations that should be considered. This study focused on looking at movements where participants were instructed to move primarily within a single anatomical reference plane. Accuracy may be different in movements that involve more variation. Additionally, fast movement was only examined for the transverse plane which was investigated closely here given its association with both performance and injury. A final limitation to consider is that each trial was short in duration (average trial length was 37.8 ± 4.5 s with the movement occurring for 6.6 ± 2.8 s) which meant error due to signal drift could not be examined.

Conclusion

Our results indicate that the relative angles computed using the sensors are valid. There was high agreement between the IMU relative angles and gold standard angles that were computed using an optoelectronic system. This indicates that the methods employed and the sensors themselves produce data that are highly accurate. While the angle data are highly accurate, agreement was lower between the IMU relative angles and those computed using the traditional anatomical marker models that are commonly employed in post-processing optoelectronic data.

Sports scientists using IMUs in-field to measure pelvis and torso dynamics, must remember IMUs do not replicate relative angles that are computed using anatomical marker models within a global coordinate system. The findings of this study suggest there are differences in the accuracy of data depending on the sensor location and this should be considered when positioning IMUs. Importantly, it is not recommended that IMUs be used to measure frontal and sagittal plane rotation of the torso relative to the pelvis or sagittal plane rotation of the upper torso relative to the pelvis as high levels of disagreement were observed here (RMSE% > 10%). Further work is needed to develop new methods to calculate anatomically valid relative angles using only IMUs mounted on the skin. We suggest that it might be necessary to combine information from multiple IMUs in a more sophisticated statistical model.

Disclosure statement

The authors report no conflict of interest.

Acknowledgement

The authors wish to thank the participants and the members of the Movement Science Discipline of the Australian Institute for their assistance with the data collection and for the provision of the equipment.

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Tables

Table 1: Comparison between the relative angles computed from the IMUs and a) marker triads surrounding the IMUs (gold standard data), b) the modelled torso-pelvis relative angles and, c) modelled thorax-pelvis relative angles for the slow rotation trials and all three anatomical reference planes. Standard deviations are indicated in brackets.

		Transverse plane rotation				Sagittal plane rotation				Frontal plane rotation			
		C7	T2	T7	Sternum	C7	T2	T7	Sternum	C7	T2	T7	Sternum
(a) Marker triads	r^2	0.99 (0.01)	0.99 (0.01)	0.97 (0.04)	0.99 (0.01)	0.98 (0.01)	0.98 (0.02)	0.98 (0.02)	0.98 (0.02)	1.00 (0.00)	1.00 (0.00)	1.00 (0.00)	1.00 (0.00)
	RMSE (°)	2.6 (1.2)	2.3 (0.7)	2.3 (0.6)	1.8 (0.7)	2.9 (1.3)	3.1 (1.5)	3.0 (1.5)	3.1 (1.2)	1.1 (0.6)	1.1 (0.5)	1.2 (0.6)	1.0 (0.4)
	RMSE%	3.2 (1.5)	3.3 (1.4)	5.1 (2.2)	2.1 (0.8)	5.6 (2.2)	5.6 (2.6)	5.7 (2.3)	6.8 (2.5)	1.4 (0.7)	1.6 (0.9)	1.8 (0.9)	1.3 (0.7)
	Bias (°)	1.8	1.5	0.8	0.4	-0.4	-0.9	-0.6	-0.5	-0.1	-0.1	-0.3	-0.2
	LOA_{upper} (°)	4.2	3.3	3.3	3.1	2.6	2.4	2.4	2.4	1.0	1.2	1.0	0.9
	LOA_{lower} (°)	-0.7	-0.4	-1.8	-2.2	-3.4	-4.3	-3.6	-3.4	-1.1	-1.4	-1.6	-1.3
(b) Thorax	r^2	0.98 (0.01)	0.98 (0.01)	0.95 (0.04)	0.99 (0.01)	0.97 (0.03)	0.97 (0.02)	0.97 (0.02)	0.97 (0.03)	0.99 (0.00)	1.00 (0.00)	1.00 (0.00)	1.00 (0.00)
	RMSE (°)	4.6 (3.3)	7.0 (2.0)	10.0 (2.7)	4.0 (1.5)	24.2 (9.6)	11.7 (7.7)	8.6 (4.9)	31.7 (11.8)	3.2 (2.0)	2.8 (2.2)	3.8 (2.0)	3.1 (1.5)
	RMSE%	5.3 (3.8)	5.7 (2.0)	11.2 (3.2)	4.7 (1.9)	40.6 (18.4)	20.8 (16.0)	16.6 (13.3)	56.8 (30.3)	4.1 (2.8)	3.8 (3.6)	5.1 (3.3)	4.0 (1.7)
	Bias (°)	1.4	1.4	-0.6	-0.2	-24.0	-8.5	4.0	31.6	0.2	0.6	-0.5	-0.7
	LOA_{upper} (°)	11.0	8.2	10.3	7.8	-4.5	13.7	21.9	55.1	7.0	7.2	6.9	5.7
	LOA_{lower} (°)	-8.1	-5.4	-11.4	-8.3	-43.5	-30.7	-13.9	8.0	-6.6	-6.0	-8.0	-7.1
(c) Torso	r^2	0.99 (0.00)	0.99 (0.01)	0.96 (0.04)	0.99 (0.00)	0.89 (0.10)	0.91 (0.07)	0.93 (0.07)	0.89 (0.09)	0.98 (0.10)	0.98 (0.01)	0.99 (0.00)	0.98 (0.01)
	RMSE (°)	5.0 (1.7)	3.8 (1.4)	6.9 (2.5)	4.2 (1.6)	29.1 (13.0)	16.1 (11.0)	11.4 (6.0)	28.5 (14.4)	10.9 (3.0)	9.4 (2.7)	7.9 (2.2)	9.2 (2.9)
	RMSE%	6.7 (1.8)	5.3 (2.1)	9.0 (3.8)	5.8 (2.1)	67.2 (34.7)	37.3 (28.8)	26.0 (15.8)	63.4 (33.7)	28.8 (8.0)	24.9 (8.1)	20.9 (6.2)	24.0 (6.7)
	Bias (°)	1.4	1.3	-0.6	-0.3	-28.1	-12.2	0.3	27.9	1.7	2.2	1.0	0.9
	LOA_{upper} (°)	9.2	6.7	7.0	6.7	-0.3	16.4	24.9	57.7	9.5	7.4	7.2	6.8
	LOA_{lower} (°)	-6.5	-4.1	-8.2	-7.2	-55.8	-40.8	-24.4	-2.0	-6.1	-3.1	-5.2	-5.1

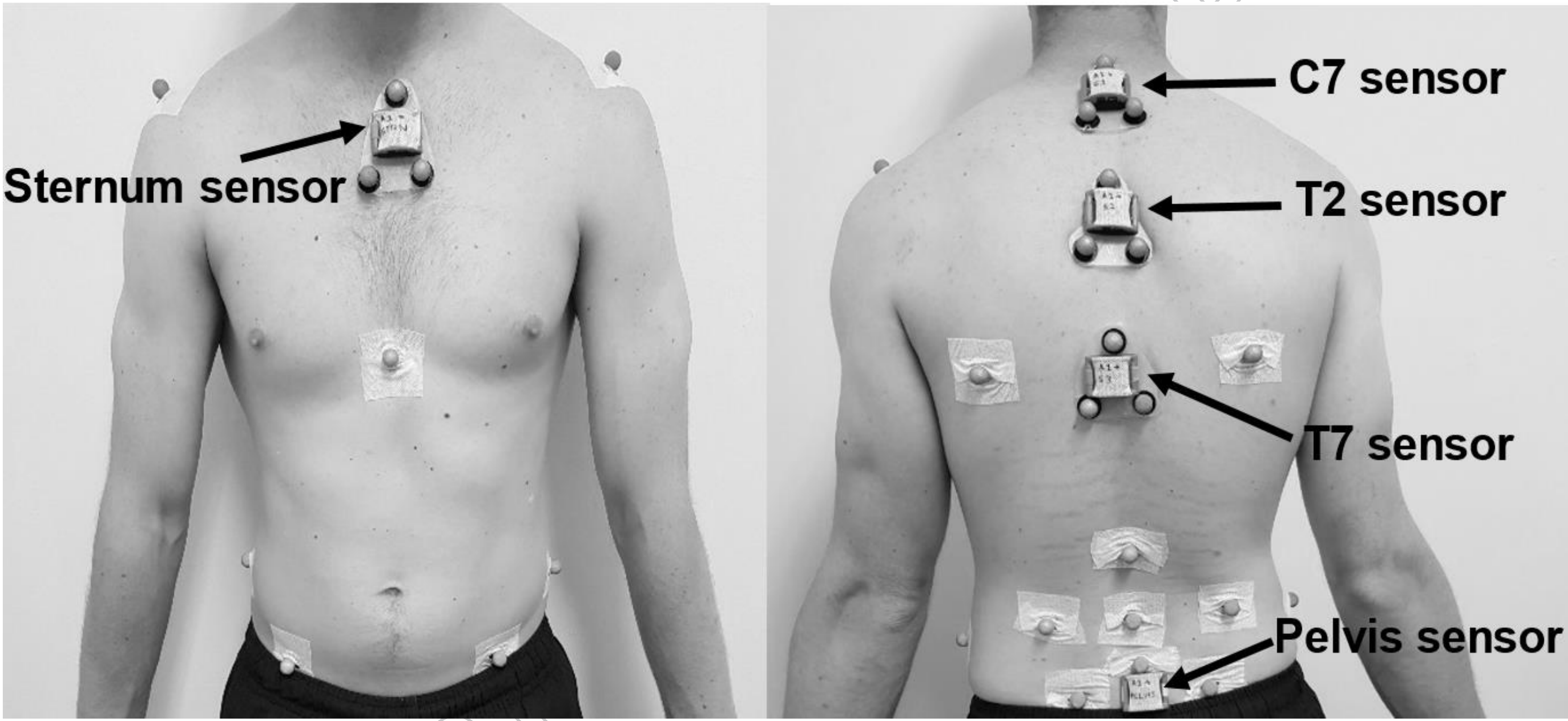
Table 2: Comparison between the relative angles computed from the IMUs and a) marker triads surrounding the IMUs (gold standard data), b) the modelled torso-pelvis relative angles and, c) modelled thorax-pelvis relative angles for the fast rotation trials in the transverse plane. Standard deviations are indicated in brackets.

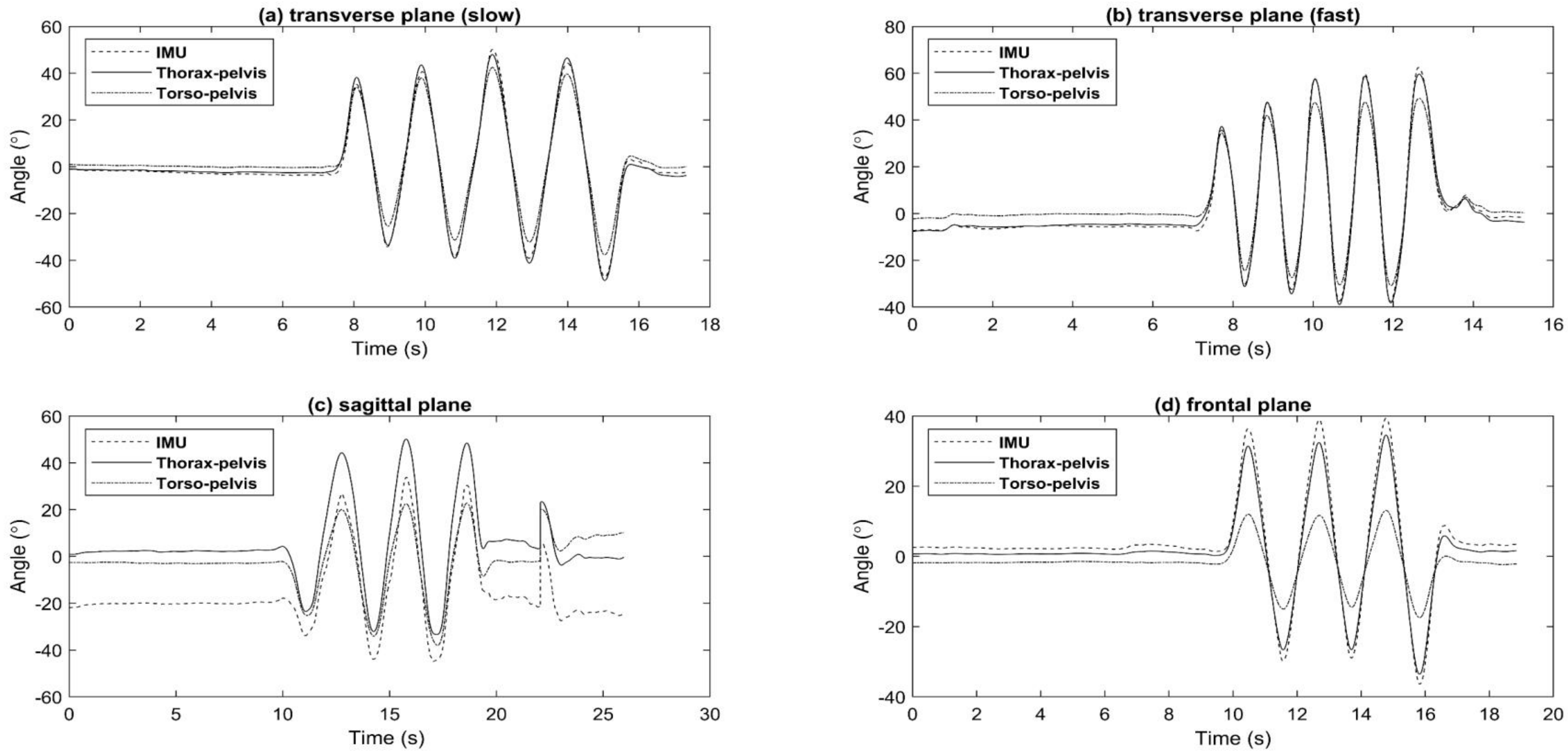
		Three fast rotations				Two fast rotations with reset between			
		C7	T2	T7	Sternum	C7	T2	T7	Sternum
(a) Marker triads	r^2	0.98 (0.01)	0.99 (0.02)	0.97 (0.03)	0.99 (0.01)	0.98 (0.01)	0.98 (0.02)	0.97 (0.02)	0.99 (0.01)
	RMSE (°)	3.3 (1.5)	2.7 (1.0)	2.9 (1.2)	2.2 (1.3)	2.2 (1.0)	2.4 (1.1)	2.4 (1.0)	2.0 (0.9)
	RMSE%	3.3 (1.6)	3.2 (1.4)	4.5 (2.3)	2.3 (1.4)	2.4 (1.3)	2.8 (1.2)	3.9 (1.8)	2.1 (1.0)
	Bias (°)	1.1	1.0	1.2	0.5	1.2	1.0	1.1	1.0
	LOA_{upper} (°)	4.6	3.1	3.4	3.5	3.2	2.9	3.1	3.5
	LOA_{lower} (°)	-2.4	-1.0	-1.1	-2.6	-0.9	-0.9	-1.0	-1.6
(b) Thorax	r^2	0.99 (0.01)	0.99 (0.01)	0.96 (0.03)	0.99 (0.01)	0.97 (0.03)	0.97 (0.03)	0.94 (0.04)	0.99 (0.01)
	RMSE (°)	3.7 (2.3)	3.8 (1.2)	7.6 (2.6)	3.7 (1.9)	3.8 (2.3)	4.3 (1.4)	6.3 (2.6)	3.6 (1.5)
	RMSE%	4.1 (2.1)	4.2 (1.2)	7.9 (1.8)	4.0 (1.8)	3.8 (1.9)	4.3 (1.4)	6.0 (2.1)	3.6 (1.7)
	Bias (°)	1.8	0.4	0.0	-0.8	0.4	0.9	0.9	-0.2
	LOA_{upper} (°)	9.9	7.6	8.7	6.4	7.7	6.8	8.5	6.7
	LOA_{lower} (°)	-6.4	-6.8	-8.7	-8.0	-7.0	-5.1	-6.8	-7.1
(c) Torso	r^2	0.99 (0.02)	0.99 (0.01)	0.97 (0.02)	0.99 (0.00)	0.97 (0.02)	0.96 (0.03)	0.95 (0.03)	0.98 (0.01)
	RMSE (°)	5.5 (1.6)	4.2 (1.8)	5.4 (1.8)	5.2 (1.5)	5.3 (2.1)	3.8 (2.4)	4.4 (1.8)	4.4 (1.6)
	RMSE%	6.6 (2.0)	5.1 (2.3)	6.0 (2.0)	5.6 (2.2)	6.0 (2.1)	4.5 (2.9)	4.9 (1.9)	5.1 (2.1)
	Bias (°)	1.4	0.5	-0.5	-1.0	0.8	0.4	-0.6	-0.7
	LOA_{upper} (°)	10.5	7.9	4.5	7.4	10.2	6.7	4.6	7.1
	LOA_{lower} (°)	-7.6	-6.8	-5.4	-9.4	-8.7	-6.0	-5.7	-8.5

Figure List

Figure 1. IMU and reflective marker locations. Visible markers not described in the text were not used during data processing.

Figure 2. Example traces of IMU relative angles for one participant for the sensor located at C7 and the model thorax-pelvis and model torso-pelvis separation angles.


Sternum sensor
C7 sensor
T2 sensor
T7 sensor
Pelvis sensor



(a) transverse plane (slow)
(b) transverse plane (fast)
(c) sagittal plane
(d) frontal plane
IMU
Thorax-pelvis
Torso-pelvis
Angle (°)
Time (s)