Comparison of the optoelectronic BTS Smart system and IMU-based MyoMotion system for the assessment of gait variables

Amadeusz Bartoszek[1], Artur Struzik[1]*, Sebastian Jaroszczuk[2],
Marek Woźniewski[3], Bogdan Pietraszewski[1]

[1] Department of Biomechanics, Wroclaw University of Health and Sport Sciences, Poland.
[2] Biomechanical Analysis Laboratory, Wroclaw University of Health and Sport Sciences, Poland.
[3] Department of Rehabilitation in Internal Diseases, Wroclaw University of Health and Sport Sciences, Poland.

Purpose: Although inertial measurement unit (IMU)-based systems have been validated against optoelectronic systems for recording joint kinematics, the accuracy of each system must be evaluated, and measurements from different systems cannot be easily compared. Therefore, this study compared the joint angles recorded using the IMU-based MyoMotion system and the optoelectronic BTS Smart-DX 700 system during Nordic walking. *Methods*: The study subject, a long-time Nordic walking instructor, was assigned to walk 12 m/trial (14 trials with 5 sampled gait cycles) at a velocity preferred for Nordic walking. The trials were simultaneously recorded by both systems. The instantaneous lower (ankle, knee, hip) and upper (shoulder, elbow, wrist) limb joint angles were recorded. *Results*: The joint angles from MyoMotion were significantly larger or smaller (depending on the joint and plane) than those from BTS. *Conclusions*: Joint angles measured by MyoMotion are not interchangeable with values from BTS, and IMU-recorded values should be interpreted carefully. However, MyoMotion can still provide information about intra-individual changes based on the joint angle profiles, e.g., following Nordic walking training.

Key words: inertial measurement unit, motion analysis, Nordic walking, reliability, validity, wearable sensors

1. Introduction

As the basic element of human locomotion, correct human gait is necessary to maintain an adequate level of comfort of life [16]. Nordic walking (NW) is a kind of gait with properly designed poles that uses a special technique developed to preserve a natural, biomechanically correct gait and posture. Through gait analysis, the gait phase can be identified, the kinematic and kinetic variables of human gait events can be determined, and the musculoskeletal functions can be quantitatively evaluated [30]. However, it is challenging to accurately evaluate multiple joints of both the lower and the upper limbs in multiple planes of movement when an individual is performing a dynamic functional activity. Currently, studies related to biomechanical analyses present two possibilities for measuring data: the optoelectronic-based measurement system and the inertial measurement unit-based system.

The gold standard in motion capture is the optoelectronic measurement system (OMS) [4]. Optical motion capture systems use passive or active reflective markers and a series of fixed cameras to track the marker positions by detecting (infrared) light, and they use this detection to estimate the 3D position of these markers via time-of-flight triangulation. Despite being more accurate than other systems, their accuracy depends on the maximum number of cameras and the field of view of each camera (they must be relative to each other); the distance between the cameras and the markers; the position, number, and type of the mark-

---

ers in the field; and the motion of the markers within the capture volume [14]. Additionally, OMSs are highly sensitive to alterations in the setup, which means that the data output will be interrupted when the cameras accidentally shift or lose sight of the markers [27], [32]. There is also a relationship between camera resolution and sampling rate that affects the quality of the data obtained because of the software. Therefore, OMSs can acquire data only in a restricted area of a specialist biomechanical analysis laboratory, which results in significant practical difficulties regarding cost, portability, calibration, synchronization, labor, and set-up.

Inertial measurement units (IMUs) have been widely recognized as a means to overcome the disadvantages of existing OMSs by aiming to offer a low cost and by being portable, real-time and relatively easy to use [30]. IMUs are devices capable of measuring various kinematic variables, such as object orientation and velocity, using accelerometers, gyroscopes and magnetometers [1]. Because no cameras are needed when working with IMUs and the system is wireless, it is possible to perform experiments outside the laboratory, which is especially interesting for collecting measurements during field sports [3], [23] and those related to locomotion in patient populations [12], [20]. However, IMUs are not without flaws, e.g., drifting of the gyroscope or metallic disturbances of the magnetometer, which can contaminate the data. Different algorithms (e.g., Kalman filter) are used to adjust the measurement signals to keep each sensor's disadvantages as small as possible [30].

Even if the validity of joint kinematic variables recorded with IMU systems has been confirmed with respect to the OMS [15], [24], [34], the measurement validity of each system (different manufacturers) needs to be independently evaluated and cannot be easily compared to another IMU or OMS systems. Validity refers to the ability of a measurement tool to reflect what it is designed to measure. However, a measure cannot be valid without being reliable. Reliability refers to the reproducibility of values of a measurement in repeated trials on the same individuals. Better reliability implies better precision of single measurements and better tracking of changes in measurements [10]. The comparison of measurement systems is best performed at high velocities of movement, when the probability of measurement error is quite high. Portable measuring devices perform much better when measuring slow movements than fast ones. However, normal walking or the gait of people with movement disorders are usually slow. Therefore, we wanted to test MyoMotion under possibly fast (extreme) walking conditions. If MyoMotion is valid dur-

ing NW gait, it most probably is to be valid during slower walking.

Therefore, the aim of this study was to compare the values of joint angles during the Nordic walking gait recorded using the IMU-based MyoMotion Research measurement system (Noraxon U.S.A. Inc., Scottsdale, Arizona) and an optoelectronic BTS Smart-DX 700 measurement system (BTS Bioengineering Corp., Garbagnate Milanese MI, Italy). To our knowledge, no previous studies have compared the measurements obtained using BTS and MyoMotion. We also investigated the intrasession reliability of the joint angle values obtained from both measurement systems.

2. Materials and methods

Participant characteristics

The subject of the study was a long-time Nordic walking instructor (woman) who had a body height of 1.73 m and a body mass of 68.8 kg. The subject ($n = 1$) was recruited using the following criteria: age between 25 and 30 years; healthy with no known neurological, cardiovascular, or musculoskeletal conditions; and a minimum of 5 years of NW instructor experience. The selection of one person for testing was to avoiding measurement errors related to significantly different inter-individual variations in NW gait techniques, body length proportions, and placement of markers and sensors on the subject's body. The study was carried out in the Biomechanical Analysis Laboratory (with PN-EN ISO 9001:2009 certification) at the Wroclaw University of Health and Sport Sciences, Poland. The research project was approved by the Senate's Research Bioethics Committee, and the procedures complied with the Declaration of Helsinki regarding human experimentation. The participant agreed to participate in the study and provided written informed consent.

Subject and equipment preparation

The task in the experiment was to cover a distance of 12 meters (14 attempts with 5 gait cycles) at a velocity preferred for the NW gait style (2.1 ± 0.1 m/s, chosen by the participant). Three initial (starting gait) and three final steps (ending and braking steps) were omitted in the analysis due to the lack of proper NW gait velocity during these steps, analogous to work by Fusca et al. [8]. As it has been confirmed, gait velocity significantly influences kinematic and kinetic gait patterns [18]. Therefore, 5 central NW gait cycles per-

formed all at a similar gait velocity were included in the analysis. The trials were simultaneously recorded by the IMU-based MyoMotion model 680 (Research) measurement system (Noraxon U.S.A. Inc., Scottsdale, Arizona) and an optoelectronic BTS Smart-DX 700 measurement system (BTS Bioengineering Corp., Garbagnate Milanese MI, Italy).

To compare BTS and MyoMotion, we chose a number of variables regarding the angles in the joints calculated by the software of these systems; additionally, these variables were obtained in both systems. The following kinematic gait variables were recorded: hip flexion-extension, hip abduction-adduction, hip internal--external rotation, knee flexion-extension, ankle dorsi--plantar flexion, shoulder flexion-extension, shoulder abduction-adduction, elbow flexion-extension and wrist abduction-adduction. MyoMotion can also register ankle abduction-adduction and ankle inversion-eversion, but they cannot be directly compared with BTS foot progression. Wrist flexion-extension has been omitted because for proper NW gait technique, it should appear only in the minimum range of motion.

At the start, for BTS, 22 retroreflective markers (6 mm in diameter) were placed on the participant according to the Davis-Heel protocol [5]. Slight modifications were introduced to the protocol by adding 4 additional markers on the upper limbs (lateral side of elbow joint – humeral head and styloid process of ulna) and 4 on the NW poles. Anthropometric measurements that provided input data for the kinematic gait analysis were taken by the BTS with the participant in a standing position on two force plates (9286A, Kistler Group, Winterthur, Switzerland). Body weight was recorded using the same plates. These measurements were taken before the experiment.

The MyoMotion IMU sensors were placed at 14 segments according to the body model, as suggested in the MR3 software instructions (Noraxon Inc., Scottsdale, AZ, USA). The sensors were attached with special fixation straps (for the pelvis) and elastic straps. For the upper limbs, the sensors were placed on the upper arms (lateral attachment at the midpoint of the humerus), forearms (on the distal section of the segment where the muscle belly was minimal), hands (at mid-portion of upper hand) and C7. For the lower limbs, sensors were attached to the shoes (top of the upper surface of the foot, slightly below the ankle), shanks (frontal surface of the tibia bone), thighs (frontal attachment to the lower quadrant of the quadriceps, slightly above the kneecap at the area of the lowest muscle belly displacement during motion) and bony area of the sacrum. The location of BTS markers and IMU sensors is shown in Fig. 1. An IMU sensor

calibration for the body in a standing position was performed before measurements were collected. After the initial calibration, there were no differences in the joint angle values recorded using the BTS and MyoMotion in the standing position before the NW gait task.

Fig. 1. The location of BTS markers and IMU sensors on the subject's body and sticks

Equipment characteristic

The MyoMotion Research 3D motion analysis system was used to investigate the kinematic variables in the NW gait. The MyoMotion IMU sensors (model 610) include a 3D accelerometer, gyroscope, and magnetometer that measure the 3D rotation angles of each IMU sensor in absolute space. By placing individual IMU sensors on two neighboring body segments, one can evaluate the range of motion in the joint placed between these segments. This principle might be extended from an individual movement of a joint to simultaneous measurements of the motion of the whole body in individual major joints. The sampling frequency for the inertial sensors was set at 200 Hz. Individual sensors are identical in functionality but are identified by a unique serial number. The angular orientation of the sensor is determined by an efficient data merging algorithm (Kalman filter) combining 15 parametric sensor readings in 4 element values expressed using quaternions. These calculations take place within the sensor with an accuracy of +1 degree in the frontal and sagittal plane and +2 in the transverse plane. The

second type of sensors used during the test have an angular velocity measuring range in 3 axes at +2000°/sec with a noise level of $0.03°/\text{sec}/\sqrt{\text{Hz}}$. The internal sampling rate for angular velocity is set at 220 Hz. The acceleration measuring range is +16 g with a maximum internal sampling rate of 800 Hz and a noise level of $110 \text{ ug}/\sqrt{\text{Hz}}$. The magnetic field sensor located in the MyoMotion sensor has a range of +1.9 Gauss with a maximum internal sampling rate of 800 Hz. The dimensions of a single sensor are 37.6 mm × 52 mm × 18.1 mm (length × width × height), and the sensor mass is 34 g. The MyoMotion sensors transmit the motion of the human body directly to the MyoMotion receiver, whose module provides wireless communication with IMU sensors at 2.4 GHz and has a typical working range of 30 m. The receiver module dimensions are 100 mm × 108 mm × 25.4 mm (length × width × height), and the mass is 215 g. As mentioned above, MyoMotion is completely wireless and does not require calibration of the measurement space.

The BTS Smart-DX 700 3D motion analysis system is a system for 3D motion analysis, which, in this study, consists of six TC IR optoelectronic cameras (up to 16 TvCs for each workstation) enabling tracking of the position in the measurement space of passive markers placed on the body of the examined person in a model strictly defined by the selected test; thus, this method allows for an evaluation of the condition of the patient's musculoskeletal system based on a completed motor task. The main tools responsible for collecting data are the aforementioned IR cameras, which have a maximum acquisition frequency of 1000 Hz (in our study, we used a setup of 250 Hz) and a sensor resolution of 1400 × 1000. The accuracy of the cameras is less than 0.2 mm on a volume of 4 × 3 × 3 m. The cameras have interchangeable lenses of type C-mount with a fixed focal length of 4.5–8 mm and zoom lenses 6–12/25 mm. The LED illuminator wavelength is 850 nm. Of course, equally important for the operation of the system are markers with spatial characteristics of 0 from 3 to 20 mm, which reflect infrared light. The obtained gait data allow for the calculation of kinematic and dynamic quantities that will be used to develop a model based on the centres of the joints, whose locations are determined in relation to the external reference system of the laboratory.

Data management

Collected data were exported as .mdx files for BTS and .dat files for MyoMotion. The operation of both

systems is based on Euler angles. The BTS makes calculations based on 3 markers on a given segment to build a reference system related to a given segment. For example, if we have a minimum of two reference systems (e.g., knee-thigh and knee-lower leg), the use of Euler angles gives us the ability to determine the angle between these two segments. MyoMotion works on the same principle except that each of the sensors has its own reference system associated with the gyroscope.

For data collected using BTS, the reconstruction and auto-labelling of marker trajectories was first performed in BTS Smart Tracker. Each trial was then visually inspected, and unmarked trajectories were manually labelled. Gaps in trajectories of up to 5 samples were joined with linear interpolation filtered. The operations led to the correct 3D model of gait. In the next step, the .mdx files from BTS Smart Tracker were implemented in BTS Smart Analyzer, where custom analysis scripts and protocols were created and resulted in kinetic and kinematic data that were saved in .mdx and .xls files. For MyoMotion, there was no post-processing performed on the data provided by this system.

For each NW gait attempt (14 in total), five central strides were used for the analysis. Movement cycles were defined using data from BTS as from the first right-foot heel strike to the subsequent heel strike on the same side. A heel strike was determined as the local minimum in the anterior-posterior position of the heel relative to the sacrum. Gait cycles (as a whole) were normalized with respect to time and averaged separately for each lower limb. Then, the mean angle profiles were calculated with standard deviations. Therefore, the curve shapes were normalized with respect to time (relative time expressed as a percentage). Additionally, for each registered joint movement, the following values were determined: maximum angle (MAX), minimum angle (MIN) and range of motion (ROM).

Statistical analysis

Concurrent validity was analyzed through a Hopkins [11] spreadsheet to quantify the relationship between the practical (MyoMotion) and criterion (BTS) measures (based on angle profiles). The validity spreadsheet uses simple linear regression to derive a calibration equation, a typical error of the estimate and an r-Pearson correlation coefficient. The criterion was the dependent variable, and the practical was the predictor in a consecutive pairwise manner. The usual scale for correlation coefficients was used for interpretation of r values: 0.0–0.1 – trivial; 0.1–0.3 – small;

0.3–0.5 – moderate; 0.5–0.7 – large; 0.7–0.9 – very large; and 0.9–1 – nearly perfect. The typical error of the estimate was standardized (SEE) by dividing by the SD of the criterion. The SEE was evaluated using half the thresholds of the modified Cohen scale: <0.1 – trivial; 0.1–0.3 – small; 0.3–0.6 – moderate; 0.6–1.0 – large; 1.0–2.0 – very large; and >2.0 – extremely large. Uncertainty in the estimates was expressed as the 90% confidence limits. To complement the correlation analysis, Bland–Altman plots were used to visualize the mean of the difference (bias) and the limits of agreement (95% confidence intervals). Intrasession reliability (i.e., the reliability of the joint angle values across the 14 NW gait attempts) of the BTS and MyoMotion was assessed with the intraclass correlation coefficient (ICC) [11]. The standardized typical error (STE) was also determined. Based on the angle profiles, the means of the instantaneous angle values over the 5 gait cycles from each attempt were analyzed. The STE should be doubled to interpret its magnitude using the thresholds for interpretation of the change in the mean (<0.2 – trivial; 0.2–0.6 – small; 0.6–1.2 – moderate; 1.2–2.0 – large; 2.0–4.0 – very large; and >4.0 – extremely large) [26]. The Shapiro–Wilk and Lilliefors tests were used to examine the distribution of individual variables. It was assumed that the data are not normally distributed. Therefore, to evaluate the differences in the peak joint angles (MIN, MAX, ROM) between the values obtained from the two different measurement systems (BTS and Myo-Motion), a nonparametric Wilcoxon test was used. In all tests performed, the level of significance was set at $\alpha = 0.05$. Statistical calculations were made using the Statistica 13.3 software package (TIBCO Software Inc., Palo Alto, CA). Furthermore, the remaining calculations were made using a Microsoft Excel 2016 spreadsheet (Microsoft Corporation, Redmond, WA).

3. Results

In Figure 2, the joint angle profiles for the right side of the body obtained from BTS and MyoMotion for the ankle (dorsi-plantar flexion), knee (flexion-extension), hip (flexion-extension, abduction-adduction and internal-external rotation), shoulder (flexion-extension and abduction-adduction), elbow (flexion-extension) and wrist (abduction-adduction) during the NW gait are presented. In Figure 3, similar profiles for the left side of the body are shown. Depending on the joint and plane, positive values of the angle (direction) correspond to flexion, abduction, external rotation, and

dorsiflexion. All MyoMotion joint angle profiles presented in Figs. 2 and 3 appear to be similar to the patterns provided by BTS.

The correlation coefficients *r* (Table 1) showed nearly perfect relationships between BTS and MyoMotion instantaneous joint angle values for ankle dorsi-plantar flexion, knee flexion-extension, hip flexion-extension, hip abduction-adduction (only for right joint), shoulder flexion-extension and elbow flexion-extension (only for left joint); a very large relationship for shoulder abduction-adduction (only for right joint) and wrist abduction-adduction (only for right joint); and large relationships for hip abduction-adduction (left joint), shoulder abduction-adduction (left joint) and elbow flexion-extension (right joint). The relationships were moderate in the case of hip internal-external rotation and wrist abduction-adduction (left joint).

However, for joint angle profiles in planes other than the sagittal, the values of SEE ranged from large to extremely large (hip abduction-adduction, hip internal-external rotation, shoulder abduction-adduction and wrist abduction-adduction). When the SEE is large, the predicted *y* values are scattered widely above and below the regression line. Therefore, high bias values, some even greater than 10°, occurred (Table 1). The joint angle values obtained using MyoMotion were significantly higher or lower (depending on the joint and plane) than the joint angle values obtained using BTS (Table 1).

Bland–Altman plots are presented in Figs. 4 and 5. For any measurement system to be valid, most of the paired differences should lie within the 95% limits of agreement, while their mean can help identify whether any system underestimates or overestimates measurements relative to the criterion (bias). The results indicate that MyoMotion overestimated or underestimated the measurements of joint angles (depending on the joint and plane). In Figures 4 and 5, most of the 100 analyzed instantaneous joint angle values are within the limits of agreement.

Additionally, in Table 2, the mean values of peak joint angles (MAX, MIN, ROM) are contained. The significant differences for almost all variables from Table 2 confirm that MyoMotion overestimated or underestimated the measurements of joint angles (depending on the joint and plane).

In Table 3, the mean ICC and STE values for the intrasession reliability of the joint angles are contained. For most joint movements, the ICC values were very large or nearly perfect, with trivial or small errors. The STE was large only for BTS shoulder flexion-extension, which indicates substantial within-subject variation of the angle values for this movement.
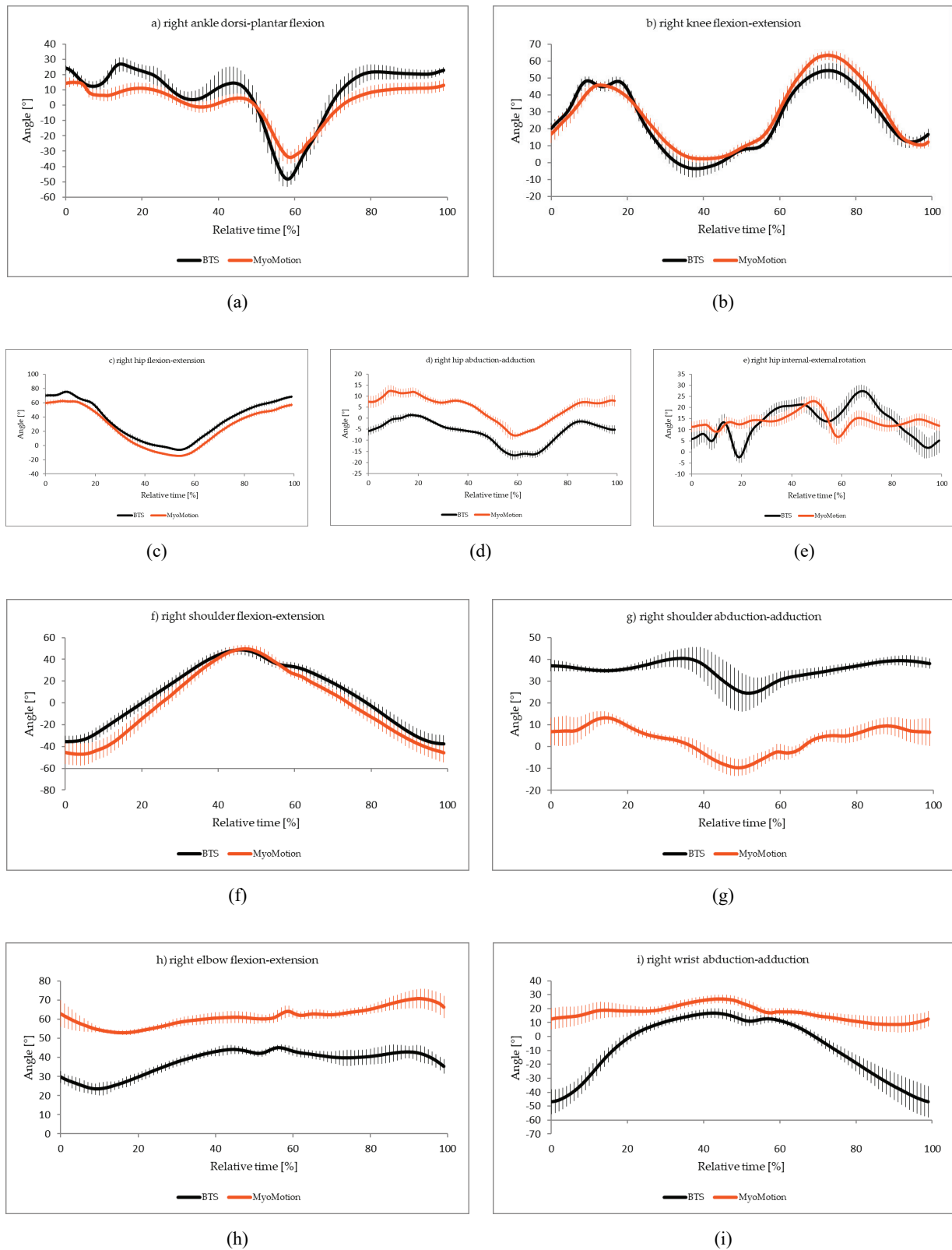
Fig. 2. Instantaneous changes (mean ± SD) of the right side of the body joint angles during NW gait
obtained from BTS (black) and MyoMotion (red) for: (a) ankle dorsi-plantar flexion,
(b) knee flexion-extension, (c) hip flexion-extension, (d) hip abduction-adduction,
(e) hip internal-external rotation, (f) shoulder flexion-extension,
(g) shoulder abduction-adduction,
(h) elbow flexion-extension, and (i) wrist abduction-adduction

Fig. 3. Instantaneous changes (mean ± SD) of the left side of the body joint angles during NW gait
obtained from BTS (black) and MyoMotion (red) for: (a) ankle dorsi-plantar flexion,
(b) knee flexion-extension, (c) hip flexion-extension, (d) hip abduction-adduction,
(e) hip internal-external rotation, (f) shoulder flexion-extension, (g) shoulder abduction-adduction,
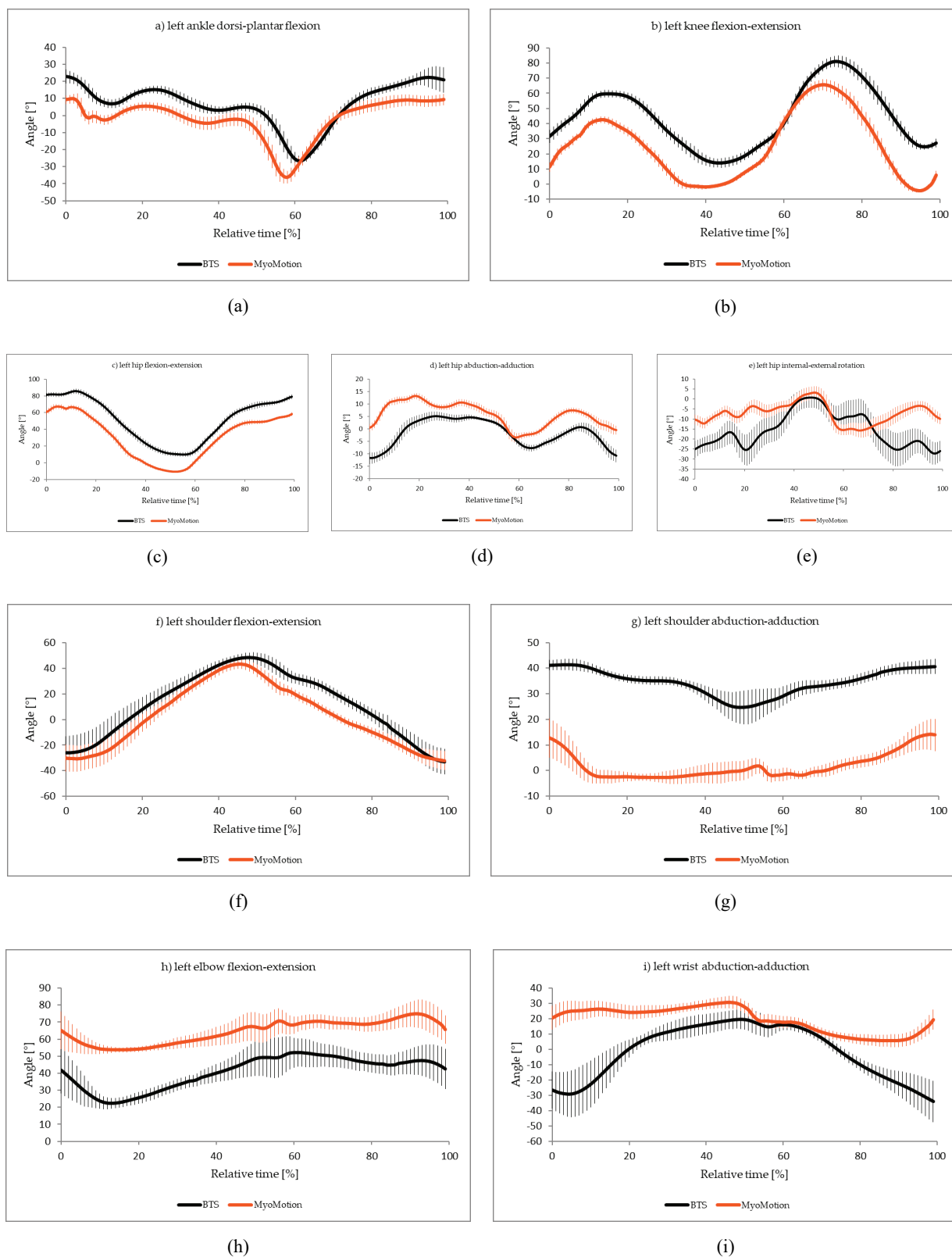(h) elbow flexion-extension and (i) wrist abduction-adduction

A. BARTOSZEK et al.

(a)

(b)

(c)

(d)

(e)

(f)

(g)

(h)

(i)
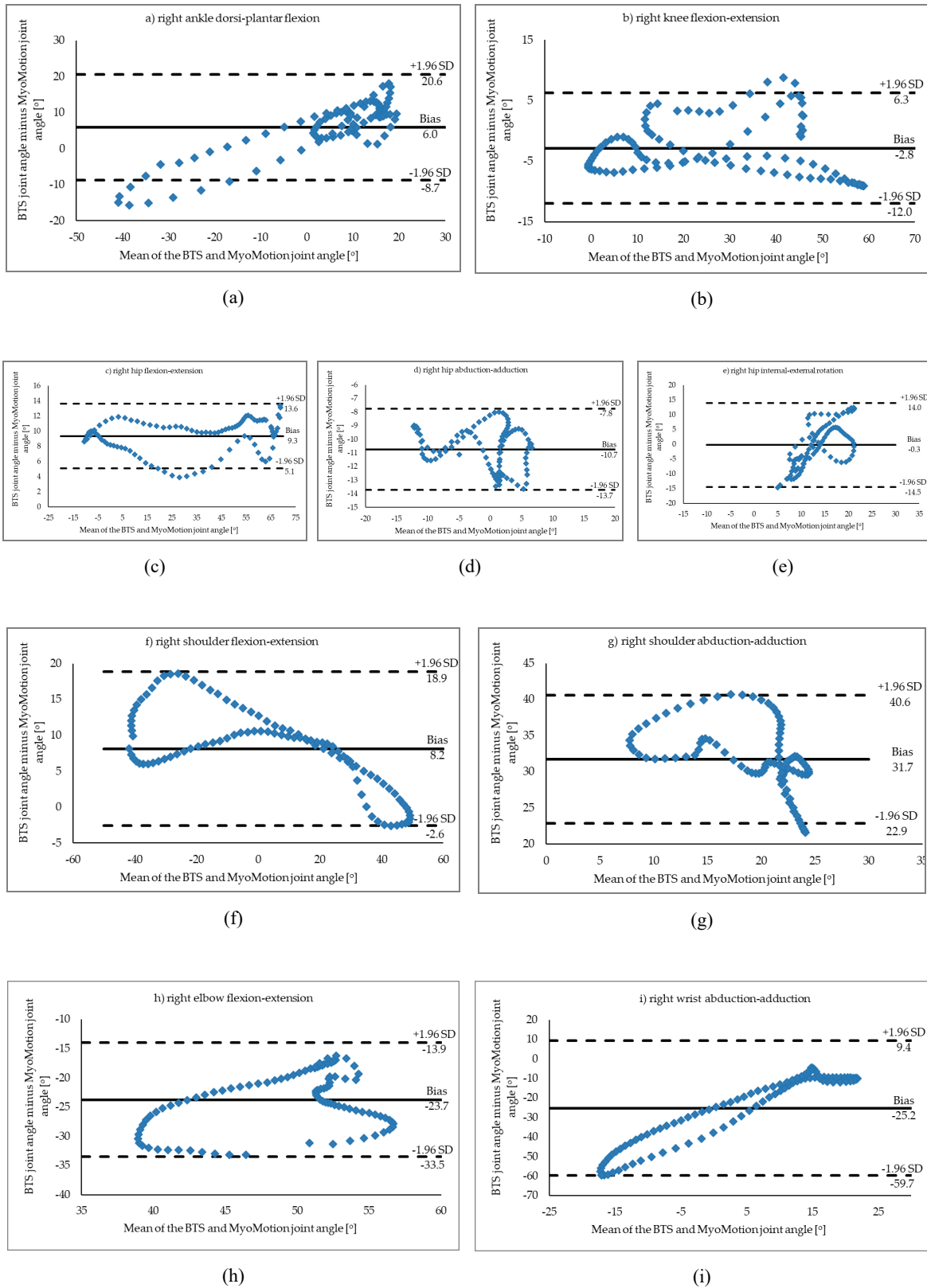
Fig. 4. Bland–Altman plots of the BTS and MyoMotion joint angles for the right side of the body:
(a) ankle dorsi-plantar flexion, (b) knee flexion-extension, (c) hip flexion-extension,
(d) hip abduction-adduction, (e) hip internal-external rotation, (f) shoulder flexion-extension,
(g) shoulder abduction-adduction, (h) elbow flexion-extension, and (i) wrist abduction-adduction

Fig. 5. Bland–Altman plots of the BTS and MyoMotion joint angles for the left side of the body:
(a) ankle dorsi-plantar flexion, (b) knee flexion-extension, (c) hip flexion-extension, (d) hip abduction-adduction,
(e) hip internal-external rotation, (f) shoulder flexion-extension, (g) shoulder abduction-adduction,
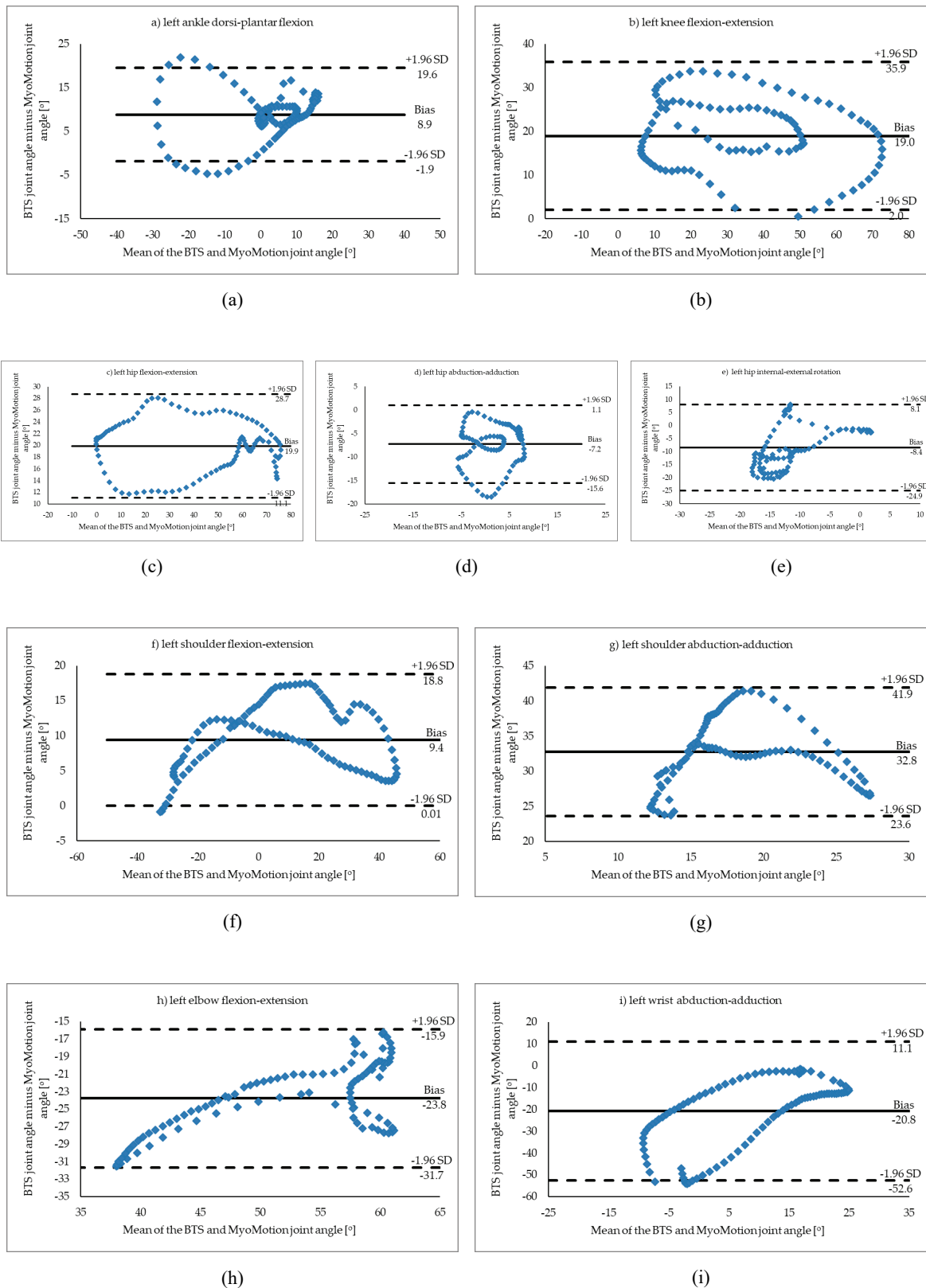(h) elbow flexion-extension, and (i) wrist abduction-adduction

Table 1. Within-rater (between systems) comparison of instantaneous joint angle values (profiles)

Joint movement	Body side	Bias [°]	SEE	r	95% LoA
Ankle flexion-extension	Right	–6.0	0.23	0.98	14.6
	Left	–8.9	0.48	0.90	10.7
Knee flexion-extension	Right	2.8	0.25	0.97	9.1
	Left	–19.0	0.43	0.92	17.0
Hip flexion-extension	Right	–9.4	0.08	1.0	4.3
	Left	–19.9	0.17	0.99	8.8
Hip abduction-adduction	Right	10.7	0.27	0.97	3.0
	Left	7.3	1.19	0.64	8.3
Hip internal-external rotation	Right	0.3	2.99	0.32	14.2
	Left	8.4	3.09	0.31	16.5
Shoulder flexion-extension	Right	–8.2	0.13	0.99	10.8
	Left	–9.4	0.19	0.98	9.4
Shoulder abduction-adduction	Right	–31.7	1.03	0.70	8.8
	Left	–32.8	1.46	0.57	9.2
Elbow flexion-extension	Right	23.7	1.12	0.67	9.8
	Left	23.8	0.39	0.93	7.9
Wrist abduction-adduction	Right	25.2	0.74	0.80	34.6
	Left	20.8	2.34	0.39	31.8

Bias – the mean of the difference between the angle values measure by the two systems; a negative (–) bias result means that BTS shows a higher value than MyoMotion; SEE – standardized error of estimate; r – Pearson correlation coefficient; 95% LoA – the 95% limits of agreement.

Table 2. Within-rater (between systems) comparison of the peak joint angle values
(MAX – maximum angle, MIN – minimum angle and ROM – range of motion)

Joint movement		Right side			Left side		
		MIN [°]	MAX [°]	ROM [°]	MIN [°]	MAX [°]	ROM [°]
Ankle flexion-extension	BTS	–50.1 ± 3.1	30.1 ± 6.0	80.2 ± 5.8	–27.2 ± 2.5	25.1 ± 5.4	52.4 ± 5.8
	MyoMotion	–36.2 ± 2.2	17.1 ± 2.8	53.2 ± 3.5	–37.5 ± 3.3	10.8 ± 2.7	48.3 ± 3.0
	Δ	13.9 ± 4.1*	–13.1 ± 6.8*	–27.0 ± 6.0*	–10.2 ± 4.3*	–14.3 ± 5.7*	–4.1 ± 4.7*
Knee flexion-extension	BTS	–4.5 ± 4.2	55.0 ± 4.6	59.4 ± 3.0	13.6 ± 3.0	81.4 ± 3.6	67.8 ± 1.9
	MyoMotion	1.7 ± 1.9	63.9 ± 2.5	62.2 ± 2.8	–4.7 ± 0.8	65.9 ± 3.4	70.6 ± 3.2
	Δ	6.2 ± 5.2*	9.0 ± 6.5*	2.7 ± 4.5*	–18.3 ± 3.2*	–15.5 ± 3.6*	2.8 ± 3.1*
Hip flexion-extension	BTS	–6.5 ± 1.1	75.8 ± 2.0	82.3 ± 2.2	9.5 ± 2.5	86.1 ± 2.6	76.7 ± 4.2
	MyoMotion	–14.9 ± 0.9	63.6 ± 2.4	78.6 ± 2.8	–11.0 ± 1.0	67.6 ± 3.0	78.5 ± 2.8
	Δ	–8.4 ± 1.6*	–12.1 ± 3.0*	–3.7 ± 3.1*	–20.4 ± 2.4*	–18.6 ± 3.2*	1.9 ± 3.5
Hip abduction-adduction	BTS	–17.6 ± 1.4	1.5 ± 1.0	19.1 ± 1.8	–12.7 ± 1.5	5.8 ± 1.3	18.5 ± 1.4
	MyoMotion	–8.3 ± 1.2	13.0 ± 2.2	21.3 ± 2.8	–3.8 ± 1.5	13.4 ± 1.1	17.2 ± 1.8
	Δ	9.3 ± 1.7*	11.5 ± 1.7*	2.2 ± 2.6*	8.9 ± 2.0*	7.5 ± 1.3*	–1.4 ± 2.0*
Hip internal-external rotation	BTS	–3.0 ± 2.8	27.8 ± 2.8	30.8 ± 2.1	–29.0 ± 5.1	1.9 ± 4.5	30.9 ± 4.8
	MyoMotion	5.1 ± 2.2	23.3 ± 2.3	18.1 ± 1.2	–15.9 ± 4.0	4.3 ± 5.1	20.2 ± 1.9
	Δ	8.1 ± 3.6*	–4.5 ± 3.4*	–12.7 ± 2.8*	13.1 ± 6.5*	2.4 ± 6.4	–10.7 ± 4.9*
Shoulder flexion-extension	BTS	–44.8 ± 6.1	44.2 ± 7.2	89.0 ± 7.8	–42.6 ± 11.1	41.4 ± 10.3	84.0 ± 11.6
	MyoMotion	–49.8 ± 8.1	50.4 ± 3.3	100.2 ± 9.4	–33.8 ± 8.8	43.5 ± 3.7	77.3 ± 10.2
	Δ	–5.0 ± 6.3	6.1 ± 5.2*	11.1 ± 6.7*	8.8 ± 10.9*	2.1 ± 10.1	–6.7 ± 8.3*
Shoulder abduction-adduction	BTS	23.7 ± 7.1	43.1 ± 4.1	19.4 ± 5.9	20.6 ± 14.6	42.8 ± 2.7	22.2 ± 16.2
	MyoMotion	–10.1 ± 3.7	13.9 ± 3.0	24.0 ± 6.1	–5.1 ± 1.7	15.8 ± 6.8	20.8 ± 6.7
	Δ	–33.8 ± 9.0*	–29.2 ± 3.0*	4.6 ± 9.1*	–25.7 ± 15.4*	–27.1 ± 7.0*	–1.4 ± 17.3
Elbow flexion-extension	BTS	22.9 ± 2.6	48.6 ± 5.3	25.7 ± 4.9	21.2 ± 2.8	57.1 ± 8.5	35.9 ± 9.0
	MyoMotion	52.8 ± 1.1	71.3 ± 4.8	18.5 ± 5.0	52.9 ± 1.9	77.5 ± 5.5	24.6 ± 4.5
	Δ	29.9 ± 2.4*	22.7 ± 6.3*	–7.2 ± 6.1*	31.7 ± 3.2*	20.3 ± 7.8*	–11.4 ± 8.3*
Wrist abduction-adduction	BTS	–51.7 ± 8.7	17.1 ± 3.5	68.8 ± 9.2	–40.5 ± 11.5	20.2 ± 6.6	60.7 ± 14.2
	MyoMotion	–27.1 ± 2.9	–7.6 ± 5.7	19.5 ± 4.7	–31.2 ± 3.6	–4.8 ± 3.5	26.4 ± 2.7
	Δ	24.6 ± 10.0*	–24.7 ± 7.4*	–49.3 ± 7.8*	9.1 ± 12.6*	–25.1 ± 8.9*	–34.3 ± 13.9*

Δ – the mean of the difference between the systems angle values; a minus (–) means that BTS shows a higher value than MyoMotion; * – significant differences.

Table 3. Intrasession reliability of the joint angle values (based on the angle profiles)
of the BTS and MyoMotion

Joint movement	Body side	BTS		MyoMotion	
		ICC	STE	ICC	STE
Ankle flexion-extension	Right	0.92	0.29	0.94	0.25
	Left	0.95	0.23	0.96	0.21
Knee flexion-extension	Right	0.97	0.18	0.97	0.18
	Left	0.98	0.14	0.98	0.14
Hip flexion-extension	Right	0.99	0.09	0.99	0.11
	Left	0.99	0.09	0.99	0.09
Hip abduction-adduction	Right	0.95	0.22	0.93	0.26
	Left	0.92	0.28	0.93	0.26
Hip internal-external rotation	Right	0.92	0.29	0.71	0.54
	Left	0.87	0.37	0.90	0.33
Shoulder flexion-extension	Right	0.98	1.41	0.98	0.15
	Left	0.82	1.35	0.98	0.13
Shoulder abduction-adduction	Right	0.73	0.52	0.80	0.45
	Left	0.81	0.44	0.68	0.56
Elbow flexion-extension	Right	0.73	0.52	0.78	0.47
	Left	0.75	0.50	0.81	0.44
Wrist abduction-adduction	Right	0.96	0.20	0.82	0.43
	Left	0.86	0.38	0.91	0.31

ICC – intraclass correlation coefficient; STE – standardized typical error.

4. Discussion

This study aimed to compare two technologically different ways of collecting data by IMU (MyoMotion) and OMS (BTS) motion analysis systems for measuring joint angles in complex motion tasks, such as the NW gait. We chose to compare OMS to the Davis–Heel protocol of gait as the "gold standard" against the IMU--based system with a conventional gait model, which is commercially implemented in Noraxon software. The BTS and MyoMotion systems showed moderate to nearly perfect relationships (Table 1) for the measurement of the instantaneous joint angle values of the upper and lower limbs. Based on the Bland–Altman plots (Figs. 4, 5), most of the 100 analyzed instantaneous joint angle values were within the limits of agreement. However, despite the good similarity in joint angle waveforms (Figs. 2, 3), the results given by the bias, SEE or calculated peak values of MIN, MAX and ROM (Table 2) had many discrepancies. The differences between the BTS and MyoMotion joint angles (bias) were often greater than 10° (Table 1). Additionally, for the joint angle profiles in planes other than the sagittal, the values of SEE were large to extremely large. Therefore, the joint angle values obtained using Myo-Motion were significantly higher or lower (depending on the joint and plane) than the joint angle values obtained using BTS. As a NW gait is a complex movement char-acterized by high velocity reaching up to 2.1 m/s and based on the literature [9], [13], [22], it was recognized that, if the discrepancy between the measurements was within ±10°, the measurement was still valid. However, the results related to the range of the compliance interval and bias between BTS and MyoMotion (Table 1) raised question about this assumption.

A possible explanation for the differences in the BTS and MyoMotion joint angle values is likely attributable to differences in the biomechanical models or biomechanical definitions between the software of these two systems. Despite the fact that both system operations are based on Euler angles, BTS makes cal-culations based on 3 markers on a given segment to build a reference system related to a given segment. For example, if in this system there are a minimum of two reference systems (e.g., knee-thigh and knee-lower leg), the use of Euler angles gives us the ability to determine the angle between these two segments. MyoMotion works on the same principle, except that each of the sensors of this system has its own reference system associated with the gyroscope.

Additionally, the differences in the BTS and Myo-Motion joint angle values may be the result of different data filtering used in both systems. MyoMotion uses the Kalman filter for estimating the internal-state of a linear dynamic system from a series of noisy meas-urements. No additional filters were applied to smooth the angular waveforms recorded by MyoMotion. In

BTS, the reconstruction of marker trajectories was performed using linear interpolation. Additional signal smoothing was done according to the BTS protocol (smooth track) by using a triangular window filter, of input order (the length of the window is 2*order+1).

When measuring the wrist abduction-adduction angles, we expected that there might be differences between systems because in BTS, we used a virtual marker to measure this angle, which was created by subtracting from the calculated values of 90° resulting from the BTS method of determining the angle on four points that were analogous to the angle between two lines. However, in this case, there were moments during the NW gait when the stick hit the ground, which may have disturbed the MyoMotion sensors. In Figure 6, an example of a wrist abduction-adduction angle profile disturbance that may occur when the stick is hit hard on the ground is shown. The artefact is visible between 20 and 30% of relative time. This profile disturbance was obtained by another person who has not yet fully mastered the proper NW gait technique. Fortunately, the NW gait technique of the examined instructor did not cause such a disturbance, as seen in Figs. 2i and 3i.
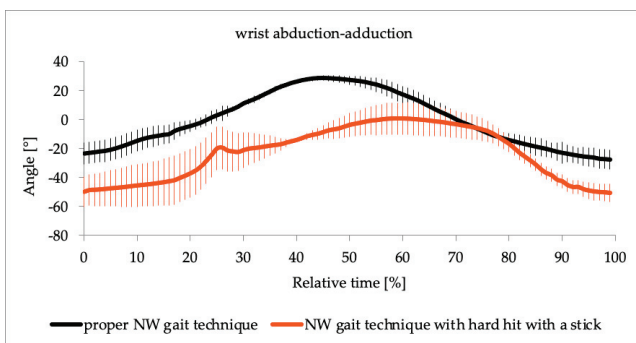

Fig. 6. Instantaneous changes (mean ± SD)
in wrist abduction-adduction angles
during proper execution of the NW gait technique (black)
and during the NW gait technique
with a hard hit with a stick (red)

The examples discussed above may have caused differences in the observed measurements. It is rather unlikely that these differences resulted from errors in sensor placement or deviation of the body position (posture) during calibration from the one required by MyoMotion or technical issues, such as magnetic field or other data transfer disruptions. However, MyoMotion sensors are attached with elastic straps that are susceptible to muscle movement and can be subject to undesirable displacement.

The biggest advantage of IMU systems over stationary laboratory devices is the possibility to use them widely in various conditions. This allows, for example, for assessing the gait-related disorders of a patient in their home or measuring during an athlete's high-velocity sport movement on a court [2], [3], [23], [28], [29]. Wearable sensing and feedback systems are portable and effective for different types of real-time human movement training and thus may be suitable for home-based or clinic-based rehabilitation applications [33]. However, the very high measurement accuracy of laboratory devices cannot be forgotten. Therefore, based on our results, the IMU systems should be used carefully when interpreting the recorded values. Both laboratory and portable devices must be valid and reliable to properly use the information they obtain [19], [25], [35].

Limitations

One limitation of this study may be the number of participants. However, this approach has strengths mentioned earlier. Additionally, the NW gait technique suggested by the International Nordic Walking Association requires specific criteria: moving the arms respecting the range of movement of natural walking, maintaining a backward pole position during the loading phase, using the poles actively and dynamically and controlling the poles by hand gripping with a grasp/release pattern. The execution of the correct NW gait technique is not obvious and automatic [17]. Different poling strategies influence the lower limb gait mechanics [31]. The motion of the lower limbs and the use of the poles often vary from those recommended [6], [21]. These deviations from the suggested NW gait technique may be caused by poor motor abilities or insufficient training, self-learning or impairments at musculoskeletal or neuromuscular level [7]. As mentioned above, deviations in the NW gait technique are noticed by both NW instructors and researchers [17]. That is why careful selection of the examined person in this study was so important so that the NW gait performed by the subject was as repetitive as possible and technically correct. Analysis of the intrasession reliability of the joint angle values (Table 3) confirmed the high repeatability of the movements performed by the examined instructor. Therefore, this analysis specifically reflects the technique of an NW expert (master's technique).

5. Conclusions

OMSs are widely considered to be more accurate, but one cannot ignore the advantages of IMU systems,

such as the low cost, portability and real-time feedback. In addition, because there is no need to use cameras and because the systems are wireless, it is possible to perform experiments outside the laboratory, which is especially applicable for performing measurements during field sports and patient locomotion. The IMU system allows the assessment of a patient's gait-related disorders in any location (e.g., at home), without the need to be in the laboratory, which is an advantage of MyoMotion. However, although there was similarity in the joint angle waveforms between the BTS and MyoMotion systems for lower and upper limb joints during the NW gait, the reported angle values were significantly different. The joint angle values obtained using MyoMotion were significantly higher or lower (depending on the joint and plane) than the joint angle values obtained using BTS. We assume that this difference is because an offset exists between these two systems, and this offset is likely a result of the different biomechanical models employed. Therefore, the MyoMotion joint angles are not interchangeable with respect to the values obtained from BTS. However, MyoMotion can still provide information about intra-individual changes based on the joint angle profiles, e.g., following NW training.

Acknowledgements

References

[1] BALASUBRAMANIAN S., ABBAS J., *Comparison of angle measurements between Vicon and MyoMotion systems*, Master's Thesis, Center for Adaptive Neural Systems (ANS), Arizona State University, 2013.

[2] CAI L., MA Y., XIONG S., ZHANG Y., *Validity and reliability of upper limb functional assessment using the microsoft kinect V2 sensor*, Appl. Bionics Biomech., 2019, 14 (1), 7175240.

[3] CAMOMILLA V., BERGAMINI E., FANTOZZI S., VANNOZZI G., *Trends supporting the in-field use of wearable inertial sensors for sport performance evaluation: a systematic review*, Sensors (Basel), 2018, 18 (3), E873.

[4] CORAZZA S., MÜNDERMANN L., GAMBARETTO E., FERRIGNO G., ANDRIACCHI T.P., *Markerless motion capture through visual hull, articulated icp and subject specific model generation*, Int. J. Comput. Vis., 2010, 87 (1–2), 156.

[5] DAVIS R.B., ÕUNPUU S., TYBURSKI D., GAGE J.R., *A gait analysis data collection and reduction technique*, Hum. Mov. Sci., 1991, 10 (5), 575–587.

[6] FIGARD-FABRE H., FABRE N., LEONARDI A., SCHENA F., *Physiological and perceptual responses to Nordic walking in obese middle-aged women in comparison with the normal walk*, Eur. J. Appl. Physiol., 2010, 108 (6), 1141–1151.

[7] FRITSCHI J.O., BROWN W.J., VAN UFFELEN J.G.Z., *On your feet: protocol for a randomized controlled trial to compare the effects of pole walking and regular walking on physical and psychosocial health in older adults*, BMC Public Health, 2014, 14 (1), 375.

[8] FUSCA M., NEGRINI F., PEREGO P., MAGONI L., MOLTENI F., ANDREONI G., *Validation of a wearable IMU system for gait analysis: protocol and application to a new system*, Appl. Sci., 2018, 8 (7), 1167.

[9] GRIBBLE P., HERTEL J., DENEGAR C., BUCKLEY W.E., *Reliability and validity of a 2-D video digitizing system during a static and a dynamic task*, J. Sport Rehabil., 2005, 14 (2), 137–149.

[10] HOPKINS W.G., *Measures of reliability in sports medicine and science*, Sports Med., 2000, 30 (1), 1–15.

[11] HOPKINS W.G., *Spreadsheets for analysis of validity and reliability*, Sportscience, 2017, 19, 36–44.

[12] HSU W.C., SUGIARTO T., LIN Y.J., YANG F.C., LIN Z.Y., SUN C.T., HSU C.L., CHOU K.N., *Multiple-wearable-sensor-based gait classification and analysis in patients with neurological disorders*, Sensors (Basel), 2018, 18 (10), 3397.

[13] HUBER M.E., SEITZ A.L., LEESER M., STERNAD D., *Validity and reliability of Kinect skeleton for measuring shoulder joint angles: a feasibility study*, Physiotherapy, 2015, 101 (4), 389–393.

[14] MALETSKY L.P., SUN J., MORTON N.A., *Accuracy of an optical active-marker system to track the relative motion of rigid bodies*, J. Biomech., 2007, 40 (3), 682–685.

[15] MUNDT M., WISSER A., DAVID S., DUPRÉ T., QUACK V., BAMER F., TINGART M., POTTHAST W., MARKERT B., *The influence of motion tasks on the accuracy of kinematic motion patterns of an imu-based measurement system*, ISBS Proc. Arch., 2017, 35 (1), 245.

[16] PALMIERI B., VADALÀ M., LAURINO C., *The FIT therapy for the treatment of musculoskeletal and neurological disorders related symptoms: a retrospective observational study*, Asian J. Med. Sci., 2019, 10 (5), 6–12.

[17] PELLEGRINI B., BOCCIA G., ZOPPIROLLI C., ROSA R., STELLA F., BORTOLAN L., RAINOLDI A., SCHENA F., *Muscular and metabolic responses to different Nordic walking techniques, when style matters*, PLoS One, 2018, 13 (4), e0195438.

[18] PIETRASZEWSKI B., WOŹNIEWSKI M., JASIŃSKI R., STRUZIK A., SZUBA A., *Changes in gait variables in patients with intermittent claudication*, BioMed. Res. Int., 2019, 7276865.

[19] PUEO B., JIMENEZ-OLMEDO J.M., LIPINSKA P., BUSKO K., PENICHET-TOMAS A., *Concurrent validity and reliability of proprietary and open-source jump mat systems for the assessment of vertical jumps in sport sciences*, Acta Bioeng. Biomech., 2018, 20 (4), 51–57.

[20] QIU S., LIU L., ZHAO H., WANG Z., JIANG Y., *MEMS inertial sensors based gait analysis for rehabilitation assessment via multi-sensor fusion*, Micromachines (Basel), 2018, 9 (9), 442.

[21] REUTER I., MEHNERT S., LEONE P., KAPS M., OECHSNER M., ENGELHARDT M., *Effects of a flexibility and relaxation programme, walking, and nordic walking on Parkinson's disease*, J. Aging. Res., 2011, 232473.

[22] RICHARDS J.G., *The measurement of human motion: a comparison of commercially available systems*, Hum. Mov. Sci., 1999, 18 (5), 589–602.

[23] Roell M., Mahler H., Lienhard J., Gehring D., Gollhofer A., Roecker K., *Validation of wearable sensors during team sport-specific movements in indoor environments*, Sensors (Basel), 2019, 19 (16), 3458.

[24] Seel T., Raisch J., Schauer T., *IMU-based joint angle measurement for gait analysis*, Sensors (Basel), 2014, 14 (4), 6891–6909.

[25] Slomka K.J., Sobota G., Skowronek T., Rzepko M., Czarny W., Juras G., *Evaluation of reliability and concurrent validity of two optoelectric systems used for recording maximum vertical jumping performance versus the gold standard*, Acta Bioeng. Biomech., 2017, 19 (2), 141–147.

[26] Smith T.B., Hopkins W.G., *Variability and predictability of finals times of elite rowers*, Med. Sci. Sports Exerc., 2011, 43 (11), 2155–2160.

[27] Stancic I., Supuk T.G., Panjkota A., *Design, development and evaluation of optical motion-tracking system based on active white light markers*, IET Sci. Meas. Technol., 2013, 7 (4), 206–214.

[28] Struzik A., Konieczny G., Grzesik K., Stawarz M., Winiarski S., Rokita A., *Relationship between lower limbs kinematic variables and effectiveness of sprint during maximum velocity phase*, Acta Bioeng. Biomech., 2015, 17 (4), 131–138.

[29] Struzik A., Konieczny G., Stawarz M., Grzesik K., Winiarski S., Rokita A., *Relationship between lower limb angular kinematic variables and the effectiveness of sprinting during the acceleration phase*, Appl. Bionics Biomech., 2016, 7480709.

[30] Tao W., Liu T., Zheng R., Feng H., *Gait analysis using wearable sensors*, Sensors (Basel), 2012, 12 (2), 2255–2283.

[31] Willson J., Torry M.R., Decker M.J., Kernozek T., Steadman J.R., *Effects of walking poles on lower extremity gait mechanics*, Med. Sci. Sports Exerc., 2001, 33 (1), 142–147.

[32] Windolf M., Gotzen N., Morlock M., *Systematic accuracy and precision analysis of video motion capturing systems – exemplified on the Vicon-460 system*, J. Biomech., 2008, 41 (12), 2776–2780.

[33] Xu J., Bao T., Lee U.H., Kinnaird C., Carender W., Huang Y., Sienko K.H., Shull P.B., *Configurable, wearable sensing and vibrotactile feedback system for real-time postural balance and gait training: proof-of-concept*, J. NeuroEng. Rehabil., 2017, 14 (1), 102.

[34] Yoon T.L., *Validity and reliability of an inertial measurement unit-based 3D angular measurement of shoulder joint motion*, J. Korean Phys. Ther., 2017, 29 (3), 145–151.

[35] Zawadzki J., Bober T., Siemienski A., *Validity analysis of the biodex system 3 dynamometer under static and isokinetic conditions*, Acta Bioeng. Biomech., 2010, 12 (4), 25–32.