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A Portable Inertial Sensing-based Spinal Motion Measurement System for Low Back Pain Assessment

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***Abstract*—Spinal motion measurement during dynamic conditions may help identify differences between individuals with and without low back pain (LBP). The purpose of this paper is to demonstrate the feasibility of an inertial sensing based, portable spinal motion measurement system for investigating the differences of the spinal motions between an LBP group and a healthy control group. During a fast flexion/extension test, we measured 3D angular motions of the pelvis, lumbar spine and thoracic spine of the two groups using the inertial sensing based system. Range of motions (ROM) and peak angular velocities were investigated to determine which variables have significant differences between the two groups ($p < 0.05$). Also, a logistic regression analysis was carried out to see the classifying ability of the LBP patients from controls using the proposed system. The result shows that LBP was particularly associated with significant decreases in peak velocities of the lumbar spinal extension motion, having the maximum 90% sensitivity and 80% specificity in the classification according to the regression analysis. The result demonstrates the possibility of the proposed inertial sensing-based system to be served as an efficient tool in providing an accurate and continuous measurement of the spinal kinematics.**

I. Introduction

Low back pain (LBP) is one of the most prevalent and costly problems of modern health care. Traditionally, functional assessment of spinal disorders has been carried out by means of subjective scales. However, quantitative and thus objective assessment of low back functional motion is critical to facilitate LBP treatment and rehabilitation [1]. Note that while patients with LBP typically do not exhibit obvious abnormalities in plain radiographs based on static motion conditions, abnormalities of the spine might be revealed by the spinal kinematics in 'dynamic' conditions [2]. Therefore, spinal motion measurement during dynamic conditions may help identify differences between individuals with and without LBP, which could lead to more targeted and improved treatment strategies aimed at regaining normal motion.

In regards to the technique of capturing the complex dynamic spinal motions, optical tracking via the use of skin markers is the most prevalent method (e.g., [3]). While optical tracking is sophisticated and accurate, the motion capture is confined to the controlled lab setting (i.e. *in-the-lab* limitation). As an alternative to the optical trackers, electromagnetic trackers have previously been employed (e.g., [4]), but their measurement range is also confined to a predefined capture space, which is limited by the transmitter-receiver distance. The reason why the in-the-lab limitation is critical in the spinal motion analysis is because it is difficult to perform comprehensive and thorough evaluations of spinal movement impairments within the time constraints available in hospitals and specialized labs [5]. Therefore, a portable and low cost spinal motion measurement system may have the potential to redirect the clinical assessments from the confines of the clinical settings to the real-life settings, i.e. the home where the patient's normal daily activities are actually carried out.

Recently, the use of miniature inertial sensors (i.e. accelerometers and gyroscopes) in human movement analysis has been gaining lots of attention due to their low cost, small size, and overall portability [6-7]. Particularly, inertial sensors are 'self-contained', meaning that they do not require any external sensing unit (e.g., a camera) and thus are highly ambulatory by simply attaching to the user's body. In fact, the feasibility of using inertial sensors for recording spinal motion has already been investigated in previous works [8-11]. Although these works demonstrated high applicability and accuracy of inertial sensors to spinal motion analysis, they did not present applications of their systems to actual clinical trials on subjects with back problems.

The purpose of this paper is to demonstrate application of a portable inertial sensing based system to 3D spinal motion analysis to investigate the differences of the spinal motions between an LBP group and a healthy control group. Using the proposed system, 3D angular motions of the pelvis, lumbar spine, and thoracic spine during a standing torso bending test were measured. Subsequently, they were analyzed kinematically and statistically to determine the differences of the spinal motions between the two groups, which may provide insight into the LBP effect on the spinal motions.

II. Methods

A. Test Protocol and Measurement System

We recruited ten LBP patients (the LBP group – six males and four females; mean age 43.2 ± 12.5 years; height 175.9 ± 7.1 cm; weight 73.8 ± 11.4 kg) and ten healthy people (the

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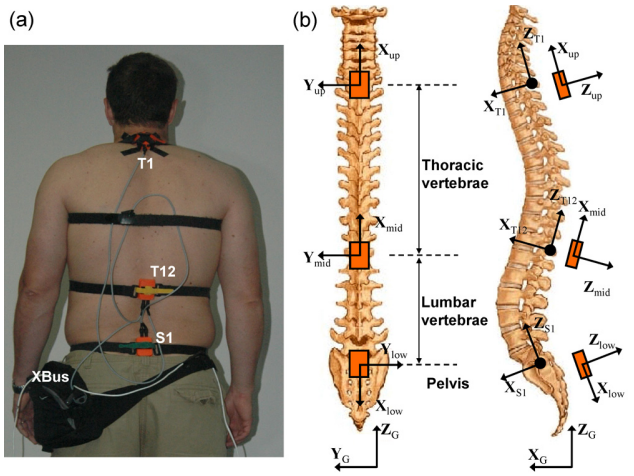

Fig. 1. **(a)** system configuration of the spinal motion measurement system comprised of three MTx sensors on T1, T12, and S1; **(b)** Segmental regions of the measurement divided by the pelvis, lumbar spine and thoracic spine, and the coordinate relationship between the sensor frames and the body frames of three vertebrae locations from the posterior view (*left*) and the left lateral view (*right*).

control group – seven males and three females; age 35.9 ± 16.6 years; height 175.2 ± 8.6 cm; weight 75.7 ± 12.3 kg). There were no significant differences between the two groups regarding age, weight and height ($p > 0.05$). Subjects in the LBP group had non-specific LBP (musculoskeletal or discogenic origin) with pain and symptoms persisting for longer than six months, but were otherwise healthy. Subjects in the control group satisfied requirements of not having a history of back pain, balance disabilities, or leg pains. This study was approved by the Office of Research Ethics of the Simon Fraser University.

The subjects were instructed to cross their arms, stand with their feet shoulder-width apart, and flex and extend their trunk repeatedly as fast as they comfortably could, i.e. fast flexion/extension test (FFE test). Previously, a similar test was used for a quantitative assessment of LBP using a 3D goniometer in [1]. Our FFE test was divided into three to include asymmetric planes of motion: FFEs (i) while maintaining the original sagittal plane (NoTwist), (ii) while maintaining maximum clockwise twist (CW), and (iii) while maintaining maximum counter clockwise twist (CCW). Ten cycles were collected for each of the FFE tests.

The spinal motion measurement system consisted of three inertial/magnetic MTx sensors (Xsens technologies B.V., Netherlands) attached onto the skin of the subject at the upper trunk (T1), middle trunk (T12) and pelvis (S1) with elastic Velcro strapping (Fig. 1.a). Each MTx consists of a tri-axial accelerometer, a tri-axial gyroscope, and a tri-axial magnetometer and provides a 3D orientation through Xsens' sensor fusion algorithm. The Velcro was tightly wrapped around the subject's body to minimize the movement between the sensor and skin. The MTx sensors were hard wired to their digital data bus system (Xsens' XBus) for data transfer which was put in a waist belt bag to be conveniently carried by the subjects. The XBus was interfaced with a laptop via a wireless Bluetooth connection at 50 Hz sampling rate.

B. Kinematic Analysis

In order to get an orientation of a body segment frame B (e.g., T1, T12, or S1) with respect to the global frame G, ${}^{G}_{B}\mathbf{R}$, the following coordinate transformation is performed:

$$ {}^{G}_{B}\mathbf{R} = {}^{G}_{F}\mathbf{R}\ {}^{F}_{S}\mathbf{R}\ {}^{S}_{B}\mathbf{R} \qquad (1) $$

where, S represents a sensor coordinate frame (e.g., up, mid, or low) which is pre-determined by the sensor manufacturer and F represents an Earth-fixed reference coordinate frame of each sensor. Fig. 1.b illustrates the above coordinate frames in our sensor setup. In (1), first, ${}^{G}_{F}\mathbf{R}$ is constant since both G and F are Earth-fixed frames that can be initially obtained through our custom automatic coordinate calibration procedure performed in a static upright standing state by each subject prior to each test. Next, ${}^{F}_{S}\mathbf{R}$ is computed by the sensor's software. Last, ${}^{S}_{B}\mathbf{R}$ is also set as constant by assuming that negligible relative orientation change of the sensors occurs with respect to the body segments. After calculating ${}^{G}_{B}\mathbf{R}$ for each segment (i.e. ${}^{G}_{T1}\mathbf{R}$, ${}^{G}_{T12}\mathbf{R}$, or ${}^{G}_{S1}\mathbf{R}$), the relative orientations of T12 with respect to S1 (i.e. ${}^{S1}_{T12}\mathbf{R}$) and of T1 with respect to T12 (i.e. ${}^{T12}_{T1}\mathbf{R}$) can be obtained (e.g., ${}^{S1}_{T12}\mathbf{R} = {}^{G}_{S1}\mathbf{R}^{T}\ {}^{G}_{T12}\mathbf{R}$), representing the postures of the lumbar and thoracic spines, respectively. Note that ${}^{G}_{S1}\mathbf{R}$ represents the posture of the pelvis with respect to the global frame. The rotation matrices, ${}^{G}_{B}\mathbf{R}$'s, are then transformed into more intuitive clinical parameters of flexion/extension, lateral bending, and axial twist, using the tilt/twist algorithm [12] as follows. First, the tilt azimuth ψ and the tilt angle ϕ need to be determined sequentially:

$$ \psi = \tan^{-1}\left(\frac{r_{23}}{r_{13}}\right) \text{ and } \phi = \tan^{-1}\left(\frac{r_{23}\sin\psi + r_{13}\cos\psi}{r_{33}}\right) \qquad (2) $$

where r_{ij} is the element in the i^{th} row and j^{th} column of the rotation matrix ${}^{G}_{B}\mathbf{R}$. Next, the flexion/extension angle FE, the lateral bending angle L, and the twist angle T are calculated from:

$$ \begin{aligned} FE &= \phi\cos\psi \\ L &= -\phi\sin\psi \\ T &= -\tan^{-1}\left(\frac{r_{31}\sin\psi - r_{32}\cos\psi}{-r_{31}\cos\psi - r_{32}\sin\psi}\right) \end{aligned} \qquad (3) $$

The angular velocity $\boldsymbol{\omega}$ of each body segment can be calculated as:

$$ \begin{cases} \boldsymbol{\omega}_{pelvis} = {}^{G}\boldsymbol{\omega}_{S1} \\ \boldsymbol{\omega}_{lumbar} = {}^{G}\boldsymbol{\omega}_{T12} - {}^{G}\boldsymbol{\omega}_{S1} \\ \boldsymbol{\omega}_{thoracic} = {}^{G}\boldsymbol{\omega}_{T1} - {}^{G}\boldsymbol{\omega}_{T12} \end{cases} \qquad (4) $$

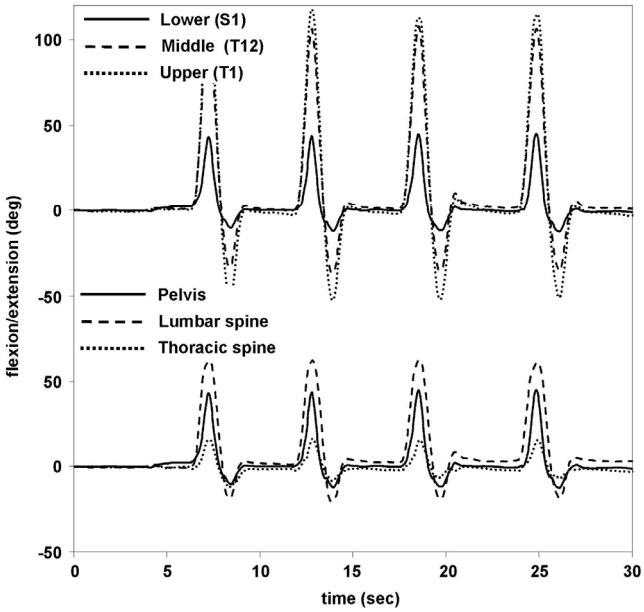

Fig. 2. Sample of the flexion-extension angles during a CCW trial of a control subject – (*upper*) S1, T12, and T1 and (*lower*) pelvis, lumbar spine, and thoracic spine.

In (4), ${}^{G}\boldsymbol{\omega}_{B}$ (i.e. ${}^{G}\boldsymbol{\omega}_{S1}$, ${}^{G}\boldsymbol{\omega}_{T12}$, or ${}^{G}\boldsymbol{\omega}_{T1}$) is obtained by the coordinate transformation, ${}^{G}\boldsymbol{\omega}_{B} = {}^{G}_{B}\mathbf{R}\,{}^{B}_{S}\mathbf{R}\,\mathbf{y}_{gyro,B}$ where $\mathbf{y}_{gyro,B}$ is the gyroscope output attached on the body *B* which is inherently expressed with respect to the sensor frame *S*.

C. Statistical Analysis

The statistical analysis has two steps: a Student's *t*-test and a binary logistic regression. The former step is to see which motion variables show a significant difference between the two groups. The latter step is to evaluate the classifying ability of the FFE tests in terms of sensitivity and specificity. Based on the results of the *t*-test, we selected input variables to the logistic regression analysis, performed by SAS statistical software v9.2 (SAS Institute Inc., NC, USA). The logistic regression model can be written as $pr = e^{z}/(1+e^{z})$ where *pr* is the probability and *z* (often referred to as the logit) is the linear combination of variables used in the model. It is defined as:

$$z = \ln\left(\frac{pr}{1-pr}\right) = \beta_0 + \beta_1 x_1 + \beta_2 x_2 + \ldots + \beta_k x_k \qquad (5)$$

where the β_0 is the intercept and $\beta_1, \beta_2, \ldots, \beta_k$ are the regression coefficients of the variables $x_1, x_2, \ldots, x_k$. In the binary logistic regression analysis, the response levels of LBP and non-LBP were set to *pr* = 1 and *pr* = 0, respectively. A forward selection procedure was used to choose the variables to be inserted into the model among the input variables. Finally, a receiver operating characteristic (ROC) plot [13] was investigated to see the sensitivity and the specificity of the binary classifier model.

TABLE I. MEANS OF ROMS AND PEAK ANGULAR VELOCITIES AND T-TEST RESULTS

Motion Variables		NoTwist	CW	CCW
ROM (°)	Flex. Pelvis	51.0 (10.0)	44.2 (6.9)	42.9 (8.5)
		58.2 (11.6)	44.8 (15.2)	45.6 (12.0)
		0.154	0.906	0.559
	Ext. Pelvis	20.1 (10.2)	17.0 (7.7)	16.1 (7.7)
		17.6 (6.3)	13.3 (3.3)	13.1 (4.1)
		0.528	0.181	0.292
	Flex. Lumbar Spine	46.7 (11.0)	41.0 (11.7)	42.2 (12.1)
		38.9 (12.0)	31.4 (12.5)	32.2 (11.5)
		0.144	0.094	0.074
	Ext. Lumbar Spine	20.4 (14.2)	14.7 (8.7)	19.0 (14.2)
		10.3 (4.5)	10.2 (7.6)	8.3 (5.1)
		0.046	0.236	**0.038**
Peak vel. (°/s)	Flex. Vel. Pelvis	108.1 (19.6)	105.1 (14.6)	101.2 (16.1)
		98.1 (35.6)	78.3 (41.8)	80.3 (39.4)
		0.446	0.072	0.139
	Ext. Vel. Pelvis	128.2 (22.5)	119.1 (18.8)	117.4 (25.9)
		104.5 (33.3)	86.7 (38.9)	84.5 (38.2)
		0.077	**0.029**	**0.037**
	Flex. Vel. Lumbar Spine	103.7 (26.6)	92.4 (27.3)	98.0 (32.0)
		69.8 (36.7)	57.8 (40.1)	59.0 (36.6)
		0.029	**0.037**	**0.021**
	Ext. Vel. Lumbar Spine	130.6 (35.1)	121.5 (36.7)	134.0 (56.8)
		80.6 (36.4)	73.5 (48.2)	70.1 (37.5)
		0.006	**0.022**	**0.008**

Means (SDs) for the control group (*upper*), the LBP group (*middle*) and the corresponding *p*-values from the *t*-test (*lower*). The bolds indicate significant differences (i.e. $p < 0.05$).

III. RESULTS

Fig. 2 shows a typical example of the flexion/extension angles during one of the FFE tests (a control subject's CCW trial). The pelvic and lumbar spinal motions were dominant while the thoracic spinal motion was relatively minimal. This indicates that the thoracic spine moved along with the lumbar spine and rarely produced its own bending motion. Therefore, subsequent analyses were focused on the pelvis and lumbar spinal regions.

Table 1 shows the means (one standard deviations) of ROMs and peak angular velocities of the pelvic and lumbar spinal flexion and extension, for both the control and LBP groups, and their corresponding *t*-test results. The *t*-test identified ten variables as statistically significant differences ($p < 0.05$) among 24 variables. The results demonstrate a greater difference between the two groups (i) in the lumbar spinal motion than the pelvic motion and (ii) in velocity variables than in ROMs. Particularly, the extension velocity of the lumbar spine had significant differences between the groups (i.e. $p < 0.01$ in NoTwist and CCW). Also, it is shown that both ROMs and velocities were decreased as the test conditions became asymmetric (i.e. the CW and CCW tests) in comparison to the symmetric test (i.e. the NoTwist test). Note that if a variable in the NoTwist test had a significant difference, the same variable in the CW and CCW tests typically showed a significant difference as well. Therefore, we could not find superior classification rates using the CW or CCW tests alone or in combination when compared to those of the NoTwist test. This implies that the NoTwist test

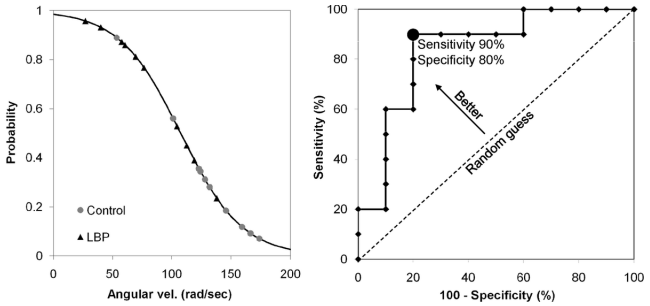

Fig. 3. The logistic function plot (*left*) and ROC plot (*right*) from the logistic regression model based on the peak extension velocity of the lumbar spine during the NoTwist FFE test.

alone (which is a relatively more comfortable test for LBP patients to perform) may provide useful classification information. This trend was observed in [1] as well.

Therefore, three variables from only the NoTwist test were selected for the logistic regression analysis – the extension ROM of the lumbar spine (p = 0.046), the peak flexion velocity of the lumbar spine (p = 0.029), and the peak extension velocity of the lumbar spine (p = 0.006). However, in the forward selection procedure of the logistic regression, only the peak extension velocity of the lumbar spine was selected because of its lowest p-value, hence the strongest classifying ability in the model. Therefore, the logistic regression model was based on the single variable having $\beta_0 = 4.144$ and $\beta_1 = -0.039$. This model had the maximum 90% sensitivity and 80% specificity (see the logistic function plot and ROC plot in Fig. 3).

IV. Discussion

In the proposed spinal motion measurement system, 3D orientations from the inertial/magnetic sensors are pivotal component of the overall measurement accuracy. The MTx sensors that we employed have an accuracy specification of 2º root mean square in dynamic motion. However, the case of the system being operated in a magnetically disturbed environment is potentially problematic as the magnetometer signals can be distorted, resulting in the heading direction errors [14]. For this reason, we performed the tests in a magnetically homogeneous environment to ensure minimal magnetic disturbances.

The results show that the low back pain is associated with significant decreases in ROMs and velocities of the pelvic/spinal motion. This may reflect an attempt by the LBP patients to reduce pain by restricting their movements of the spine, due to the presence of LBP itself or the fear-avoidance behavior associated with the LBP [15]. Particularly, the result of the regression model shows that the lumbar spinal extension velocity had the sensitivity of 90% and specificity of 80% in classifying non-specific LBP patients from controls, indicating considerable diagnostic capability. Therefore, this variable may be considered as an important measure in the rehabilitation and treatment of LBP patients, by quantitatively monitoring its improvement. This result also demonstrates the feasibility of using the proposed spinal motion measurement system for detecting classifiable differences in spinal motions between the LBP group and the control group.

In conclusion, this study proposes an ambulatory and cost effective measurement system that facilitates monitoring, recording and analysis of spinal motions during patients' normal life conditions. If quantitative measurements can be performed in patients' real-life settings for extended time periods, more effective, unbiased, and individualized spinal assessment and treatment strategies could potentially bear by the clinicians. For this purpose, the proposed inertial sensing-based spinal motion measurement system can serve as an efficient tool in providing an accurate and continuous measurement of the spinal kinematics.

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