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MEMS SENSOR APPLICATION FOR THE MOTION ANALYSIS IN SPORTS SCIENCE

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Abstract. *The author has been applied MEMS accelerometers or the piezoelectric vibrating gyroscopes for the motion analysis in sports activities by attaching them onto the athletes' body segment. As for the upper extremity dominant sports, such as swimming, the inertia sensor should be attached onto the hand or wrist joint. Then several evidences of their skill were to revealed by the sensor data. Unlike traditional cinematographical measurement method in sports science, the inertia sensor motion analysis method was given the status that its obtained data was based on not the absolute coordinate system but the movement coordinate system. The author has been proposed the swimmer's skill evaluation using their wrist tri-axial acceleration. So far, the discrimination of the swimming styles and the segmentation of the underwater stroke phases could be achieved. In addition, the physiological response was also detected on the wrist acceleation when the swimmer was fatigued in analsis methodthe intensive training situation. Adding gyroscopes to analsis methodthe accelerometers in the measurement of the sports analsis methodperformance, the detailed estimation of the atheltes' analsis methodmotion could be achieved. In addition to the swimming, the author will present MEMS sensor analsis methodapplication to the golf swing motion analysis in this analsis methodpaper The characteristics of the down swing motion on analsis methodboth the professional and amateur golfers could be obtained and analysed usign several tri-axial accelerometers and the synchronized video camra. In this paper, the analsis methodauthor also proposes an analysis method for the pattern analsis methodmatching in the sensor based motin analsis method.*

Keywords: *Sports Science, Motion Analysis, Accelerometer, Gyroscope*

1. Introduction

In the field of the sports engineering or the sports biomechanics, the videographical analysis method has been applied for the human motion analysis. However, a wide range or the long term of the motion in the outdoor circumstances prevents us from measuring whole kinematical parameters by using videographical method. The author has proposed inertia sensors such as accelerometers or gyroscopes as an alternative to the videographical motion analysis (Ohgi et al., 1999, 2000, 2003, 2002, Ohgi, 2004). Unlike the video methodology, our method using the inertia sensors is characterized by attaching sensors onto the subject body itself. Such accelerometry could not been applied for the sports science, because of the problem of sensor size or its cost. The miniaturization by MEMS technology and the progress of the impact-registent specification, the inertia sensors enable us to measure human or equipment movement in our sports activities. However, it is to be noted that we must understand the measured acceleration and the angular velocity from the body or object attached inertia sensors indicate the signal based on the local coordinate system not the global coordinate system. Especially, the acceleration from the accelerometer which was originated on the local coordinate system, namely the movement coordinate system, contains four different kinds of acceleration component, such as the gravitational, translational, centrifugal and tangential accelerations. Theoretically, we could not divided measured acceleration into each component by only a single acceleration sensor. Thus, we are using both the accelerometry and the videographical observation now, which means the three dimensional image analysis method at the same time. In this paper, the author will introduce our studies about swimming and golf research using sensor measurement method and discuss the possibility of the MEMS inertia sensors for the sports science in near future.

2. Application for Swimming Stroke Analysis

2.1 Background

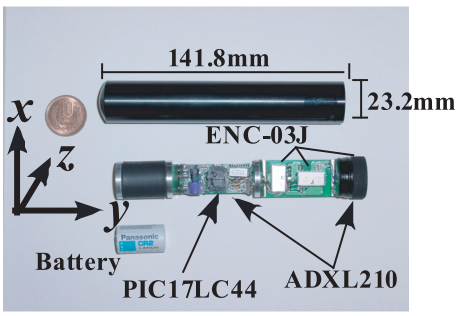
Three dimensional motion analysis method (Abdel-Aziz & Karara, 1971) has been a major approach to observe a swimmer's underwater stroke motion (Cappaert et al., 1995, Liu et al., 1993). For the three dimensional motion analysis, usually, we need more than two underwater cameras. However, this videographical method was inappropriate for the swimming research. Because it took long time and work for the digitizing process. Since the swimming stroke motion is performed underwater, we cannot use an optical or magnetic motion capture system, which has been a popular method recent years. So, the researchers could not provide the real time feedback to both the swimmers and coaches on the poolside. Therefore, another alternative has been expected for the swimming stroke analysis and swimming coaching.

2.2 Accelerometer and Gyroscope Data Logger

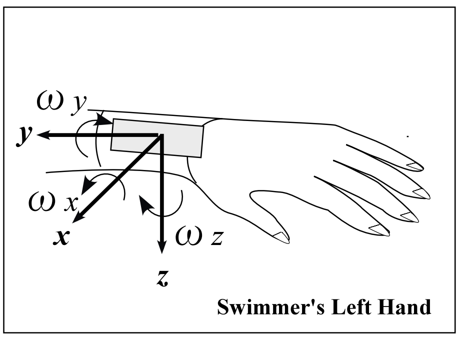
The author proposed a new method, which provides us the swimmer's stroke kinematics using their wrist acceleration and the angular velocity by the inertia sensors. The authors reported that the discrimination of the swimming stroke styles between the crawl stroke and breaststroke was achieved using the tri-axial swimmer's wrist acceleration (Ohgi et al., 1999). In addition, we also reported that the identification of the underwater stroke phase could be possible in the crawl stroke (Ohgi et al., 2000). Until those experimental studies, unfortunately, our these accelerometry studies on swimming stroke measurement were performed by wired data acquisition. Therefore, the author has been developed inertia sensor data logger device with a micro processor and the nonvolatile memory. The author developed a tri-axial accelerometer data logger (Prototype I) and a tri-axial accelerometer with tri-axial gyroscope data logger (Prototype II). The specification of the sensor data logger for the swimmer is shown in Tab. 1. Figure 1 (left) shows the developed acceleration and gyroscope data logger, Prototype II. Prototype II is able to measure the tri-axial acceleration and the angular velocity simultaneously by using two bi-axial acceleration sensor ICs (ADXL210, Analog Devices, Inc.,) and the triple gyroscopes (ENC-03J, Murata Corp.,). Also Fig. 1 shows the local coordinate system of the data logger. Figure 2 shows the signs of the angular velocity on both the x–axis and z–axis, which would be changed depend on the forearm pronation or supination configuration. From the anatomical rest position, the centrifugal component would be observed on the y–axis acceleration by the rotational movement, $A_y > 0$. Thus, in this flexion phase, the x–axis and the z–axis angular velocity changed $\omega_x < 0$, $\omega_z = 0$, respectively. From this anatomical rest position, when we pronate our forearm 90 degree with our pronation, the palm plane faces our trunk. Then, we flex our left elbow from this configuration, the y–axis acceleration also outputs centrifugal acceleration component and changes $Ay > 0$. On the other hand, as for the angular velocities, $\omega_x = 0$ and $\omega_z > 0$ would be observed. This means the sign of the angular velocity suggests the rotational orientation of the forearm during the stroke.

Table 1: Specification of the inertia sensor data logger for the swimming stroke analysis

	Prototype I	Prototype II
Dimension	88×21(mm)	141.8×23.2(mm)
Weight	50g	78g
Accelerometer	2 ×ADXL210	←
Gyroscope	–	3×ENC-03J
Processor	PIC17LC44	←
A/D	MAX147(12bit)	←
Sampling Rate	∼128Hz	←
Memory	32Mbit	128Mbit
Battery	CR1/3N	CR2
Measurement Duration	1.45h	2.9h



Photograph of Prototype II



Local coordinate system of the logger

Figure 1: Accelerometer and gyroscope data logger, Prototype II

2.3 Swmming Stroke Analysis using Wrist Kenematical Data using Sensor Data Logger

It is well-known that we have five stroke phases in our front crawl swimming. These are the entry and stretch, downsweep, insweep, upsweep and release and recovery phases (Maglischo, 1993). Among these five phases, four stroke phases except the release and recovery are performed underwater. Thus, from the view point of the fluid dynamics, these

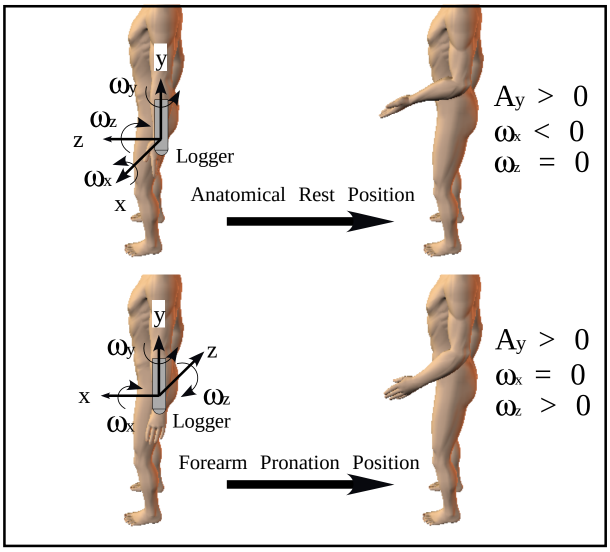


Figure 2: The sign of the wrist angular velocity in elbow flexion with its pronation or supination

underwater stroke phases are very important for the swimmers. Recently, the swimming stroke mechanics on the crawl stroke has changed drastically. Ito proposed straight arm pull motion, I-shaped pull makes maximum thrust force than that of the S-shaped pull pattern, which has been believed the best stroke solution for the swimmers in these thirty years (Ito & Okuno, 2003). It is very difficult to distinguish four underwater stroke phases and examine whether or not a swimmer performs I-shaped pull or S-shaped pull pattern, by the visual observation from above the water. In addition, if we could use underwater cameras for the stroke observation, the discrimination of the stroke phases requires the three dimensional motion analysis. It takes long time to provide the result of the stroke analysis to the swimmers and coaches. So these image based analytical feedback can not be contributed in real situation. For these reason, a new innovation, which can observe the stroke motion of the swimmer's forearm and discriminate the stroke phases has been expected by the coaching. It all comes down to the fact that the author applys inertia sensors to the swimming stroke analysis. Inertia sensors has an advantage in its prompt feedback to the swimmers and coaches on the poolside.

Figure 3 shows the experimental results of a certain college swimmer on his fast speed crawl stroke trial. The alphabetical symbols from **A** to **F** in both the kinematical data and the three dimensional hand path figure indicate the distinctive acceleration point such as the local maximum, minimum and the alternation of the acceleration signs in the single stroke. In this figure, stroke phases were determined by these distinctive acceleration pattern. Only the transition instance between the upsweep phase and release and recovery phases was determined by the three dimensional motion analysis method. We can see that there was no rotational movement for the x-axis during the stretch phase (phase **AB**) in the crawl stroke. Because the x-axis angular velocity, ω_x showed nearly $0 deg/s$. However, ω_y, changed positive to negative during the stretch phase. It means that the swimmer's forearm alternated supination to pronation in this phase. After the stretch phase, the swimmer's hand began to move downward (phase **BC**, downsweep). We can see that the angular velocity ω_y changed from negative to positive during the downsweep phase **BC**. The results showed the pronation of the swimmer's forearm at the first portion and then supination at the latter portion in the downsweep phase for all subjects. Between **C** and **D**, it was called the insweep phase, when the swimmer sculls his hand toward the trunk. At the end of the insweep phase, since the swimmer finishes his insweep motion by the shoulder abduction and the elbow flexion, the distance from his shoulder joint and the hand, namely the rotational radius would decrease steeply. Because of this motion, the centrifugal acceleration must be remarkably decreased and have a local minimum on A_y at timing **D**. At that time, the x- and y-axis angular velocities, $\omega_x = 0$, $\omega_y = 0$ were observed. And the z-axis angular velocity, ω_z changed from positive to negative after the insweep motion. In the experimental results, from the timing **D** and **F** corresponded to the upsweep phase. In the upsweep phase, the x–axis and z–axis angular velocity, ω_x and ω_z showed almost negative value. As for the y–axis angular velocity, ω_y, showed always positive (supination) for all subjects.

2.4 Comparing Kinematical Time Series Data using Dynamic Time Warping

As mentioned above, the stroke phase discrimination and the swimmer's rotational stroke motion has been revealed from our studies using the accelerometers and the gyroscope sensor data logger. Also we had proposed that the wrist acceleration changes in the swimmer's fatigue situation (Ohgi et al., 2002). The inertia sensor device will be valuable for not only the detection of the stroke change in fatigue condition, but also the effectiveness of the skill training or coaching instruction. In addition, it will be available to examine what stroke style a swimmer should chose by himself comparing

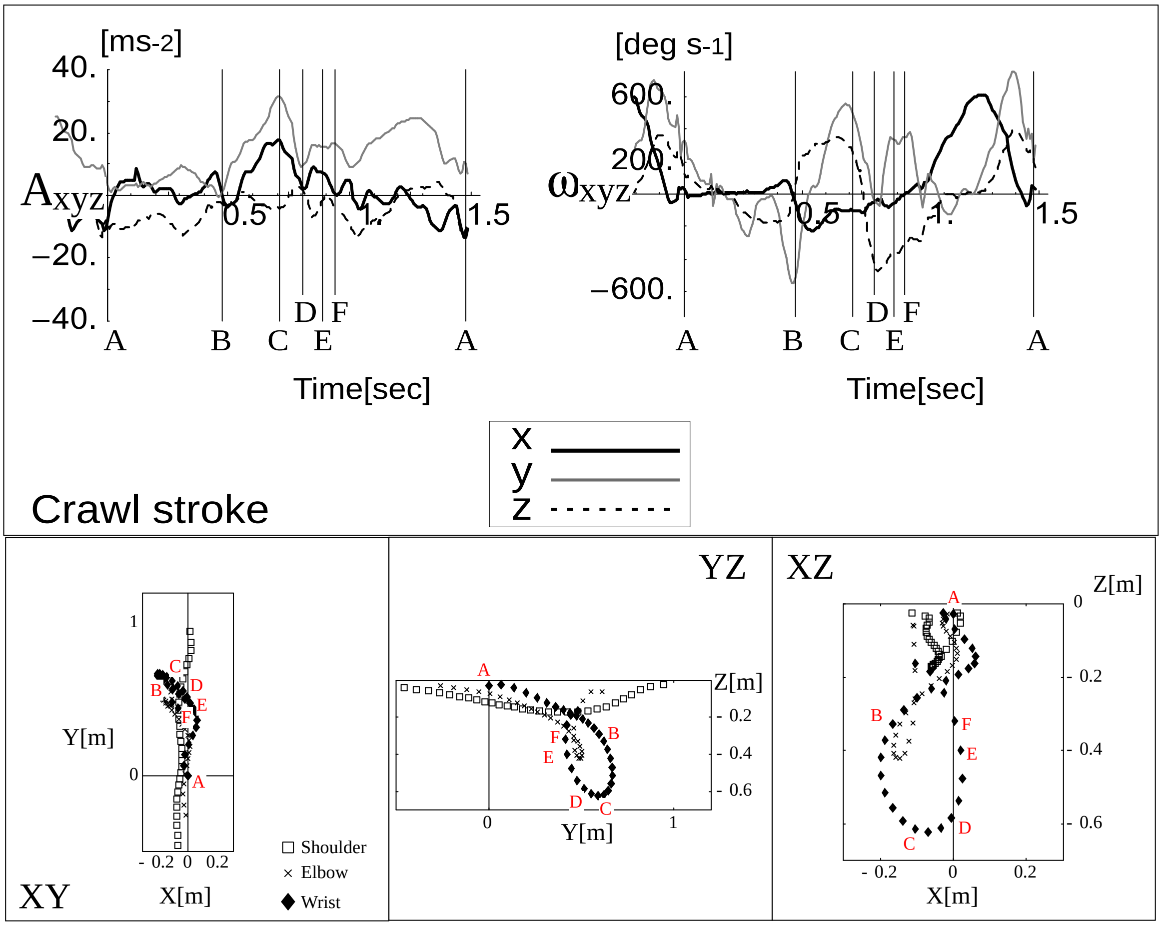


Figure 3: The swimmer's tri-axial wrist acceleration, the angular velocity and the stroke path in the crawl stroke

to his stroke kinematical data with the top athletes' one.

Because both the length and the magnitude of the sensor data can vary depend on each trial on any human motion, in the field of the biomechanics, we are usualy taking a way that we would normalize original data into 100% as its data length . However, this time normalization is not adequate for the pattern matching of the kinematical data. For this purpose, we need a method to compare the sensor time series data independent of the data length and the magnitude. The author propose the dynamic time warping (DTW) for matching the kinematical data on the human motion.

DTW is a classical and well-known speech recognition method for detecting a phoneme or a word from the different length of utterance and its vocal quality (Sakoe & Chiba, 1978). And, DTW is one of the optimization method for determining a non linear distance and coincidence between two different time series data.

Figure 4 illustrates the results of the dynamic time warping for the swimmer's wrist longitudinal acceleration (A_y) in case of two different speed crawl stroke trial on same subject. You can see the length and the magnitude of both data are different each other. Blue connecting lines between two time series, such as slow/middle (left) or slow/fast (right) trial data, are indicating the nearest corresponding matching data point between these accelerations. The DTW scores by the manhattan block distance function are 158.541m/s^2 and 261.034m/s^2 for slow/middle and slow/fast trial comparison respectively. The author expects this dynamic time warping method for the sports biomechanics or the human engineering in order to determine the matching score of different personal kinematical data.

3. Application for Golf Swing Analysis

3.1 Driver Swing Diagnosing System

The author has been developed a golf swing diagnosing system using both the accelerometers and the video camera. The measurement system is illustrated in Fig. 5. The application was developed using LabVIEW6i (National Instsu-ruments Ltd.) on the Windows OS platform. For the measurement of the golfer's trunk, left forearm and driver head motion, a low (H48B,± 29.4m/s^2, Hitachi Metals, Ltd.), middle (H48M, ±490m/s^2) and high (H48N, ±980m/s^2) capacity piezoresistive MEMS acceleration sensors were applied (Tab. 2, Fig. 6). The local coordinate system for each

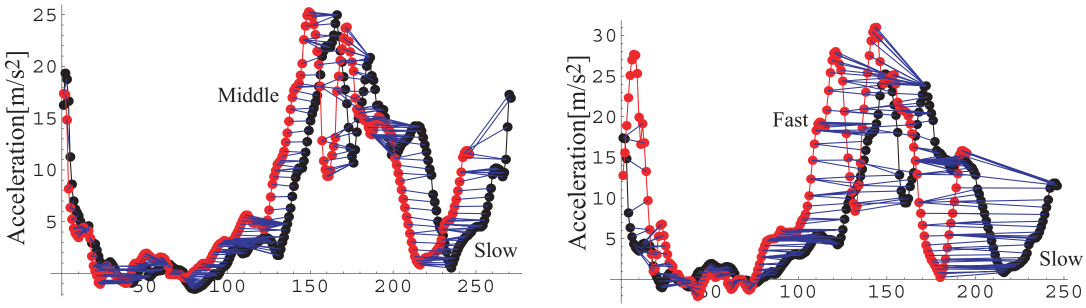


Figure 4: Acceleration comparison by the dynamic time warping. The figure shows the comparison between slow/middle speed crawl stroke trial (left) and slow/fast trial (right) using DTW. DTW scores based on the manhattan block distance function are 158.541 for slow/middle comparison and 261.034 for slow/fast comparison.

acceleration sensor was defined in Fig. 6. X, Y and Z-axis of the golfer's body at his low back was lateral, vertical and dorsoventral direction. As for the the wrist joint of the left forearm, those correspond to the radial–ulnar, the distal–proxial and the dosal–palmar direction respectively. The local coordinate system of the driver club head was defined at the address position. X, Y and Z-axis of the club head correspond to the target line direction, the proximal–distal direction with respect to the shaft and back to the front direction from the golfer's view. All acceleration signals were acquired by data acquisition board (NI–PCI6025E) at 900 Hz sampling frequency. For the videography, a synchronized progressive camera (30fps) was used and its signal was acquired by video processing board (NI–PCI1408). Forty proximate and ten immediate video frames around the impact were saved as jpeg data files. Five professional and three amateur golfers involved in the experiment. Each golfer was directed in swing normaly as possible as he can with a sensored driver.

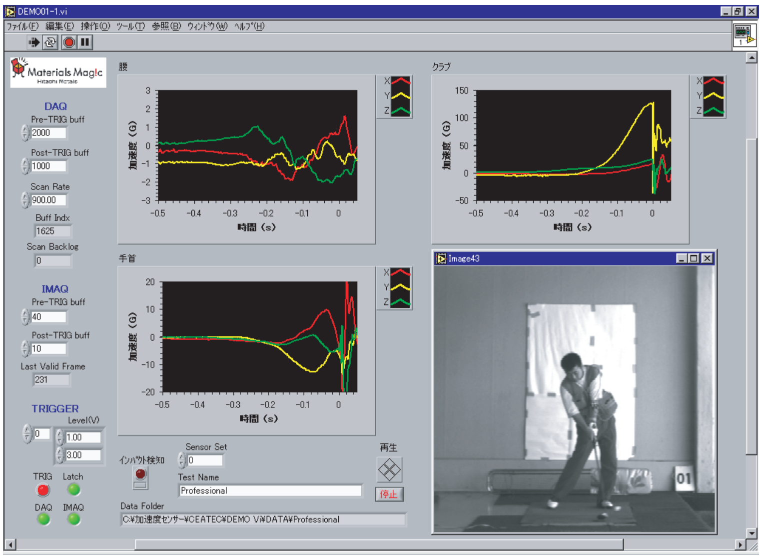
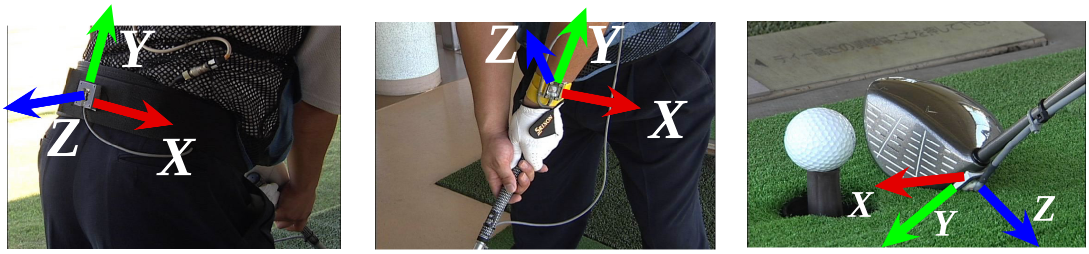


Figure 5: Golf swing diagnosing application

Table 2: Acceleration sensors for the golf swing experiment

	Acceleration Sensor	Capacity
Low back L5	H48B	±3G
Left wrist	H48M	±50G
Driver club head	H48N	±200G*



(A) Trunk (B) Wrist (C) Club head

Figure 6: Acceleration sensor for the golf swing experiment

3.2 Golfer's Body Motion in the Driver Swing

An example of the experimental results is shown in Fig. 7. In each figure, the X, Y and Z-axis acceleration indicates in red, green and blue line respectively. When the down swing begun on all professional golfers, X-axis acceleration at the hip decreased and kept negative value for a while. Then, –70ms to –30ms (**C**) before the impact , it changed to positive. The reason why X-axis acceleration of the hip changed to positive value that the professional golfers might to decelerate backward direction in order to prevent their body to move forward. However, as for the amateur golfers, we can see that the transition timing from negative to positive varied depend on their each trial, and it varied between –40ms to –15ms. In some cases in sub.H (amateur), this transition approximately appreared at the impact (Fig. 7). According to Jorgensen (Jorgensen, 1996), such delay of the golfer's deceleration of sway motion is likely to cause insufficient energy transfer from the trunk to both the upper extremity and the club. Off course, such extreme delay that a transisiton timing occured near or after the impact is undesirable. Because it means that the golfer's whole body will move forward.

The wrist acceleration sensor was attached onto the distal end of the forearm. The Y-axis acceleration of the wrist which was along with the longitudinal axis, had a centrifugal acceleration as a major component. Inoue et al. (Inoue, 1997, Inoue et al., 1999, 2000), Umegaki et al. (Umegaki et al., 1998), Kobayashi, et al. (Kobayashi et al., 1997) argued that the uncock leads the kinetic energy to transfer to the golf club in the down swing. Thus, the rotational velocity of the upper extremities decreased steeply in consequence. Based on their statements, Y-axis acceleration at the left wrist also decreased. Because, under our configuration of the local coordinate system, the centrifugal acceleration indicates negative value. Therefore, a decrease of the rotational angular velocity of the forearm must cause the increase of the Y-axis wrist acceleration. Actually, as for all professional golfers, their wrist Y-axis acceleration during the down swing phase had a local minimum at –100ms (**B**) before the impact. In contrast with the professional golfers, sub.O who is an amateur golfer handicapped under 20, his Y-axis acceleration of the wrist decreased until the impact. This means that his uncock operation was unexcuted so that his upper extremity and the club were rotated with combined. It is suggested that the wrist joint torque was likely to be kept even after the appropriate uncocking timing in his down swing phase. The most important skill of this energy transfer is that the uncock must be carried out appropriate timing. Therefore, the local minimum of the Y-axis acceleration at the wrist could be used for the indication of the release of wrist cocking. According to common features among professional golfers, the appropriate swing skill, especially, the time sequence of the swing motion should be as follows, (1) First of all, the golfer's trunk starts to move, which means the lateral sway motion to the target direction, then changes come out with their hip X and Z-axis acceleration. (2) Because his trunk accelerates to the direction of his target line, his hip X-axis acceleration decreases. (3) Following to that, because his upper extremities starts to swing, his wrist Y-axis acceleration begins to decrease. (4) About –100ms before the impact, his wrist cock releases and the local minimum might be seen on the wrist Y-axis acceleration. (5) The opposite lateral sway occurs –70ms to –30ms before the impact. Thus, the X and Z-axis acceleration changes negative to positive. (6)Impact completes. When such sequence of acceleration pattern observed, he would be highly skilled.

3.3 Variability of the Swing Motion

Furthermore, in fact, the stability of the repeated swing should be important. From Fig. 8, we can see that the within subject variability on the subjects. The professional golfer sub.F has small variability through the whole swing phase. As for the amateur golfer, sub.H was already pointed out that he could not perform his lateral sway. His within subject variability data encourages to that indication. The variability of his hip X-axis acceleration at the latter portion during the down swing was relatively larger than other phase. At that time, the variability also can be seen in his wrist acceleration. Therefore, he was unskilled and unstable on both the trunk and the arm motion during the latter phase in down swing. As for sub.O, who was pointed out that he tended to be late to release his uncock, would be unskilled and unstable on his uncock motion. In this way, our developed application can be used with an aim to determine golfer's swing variability.

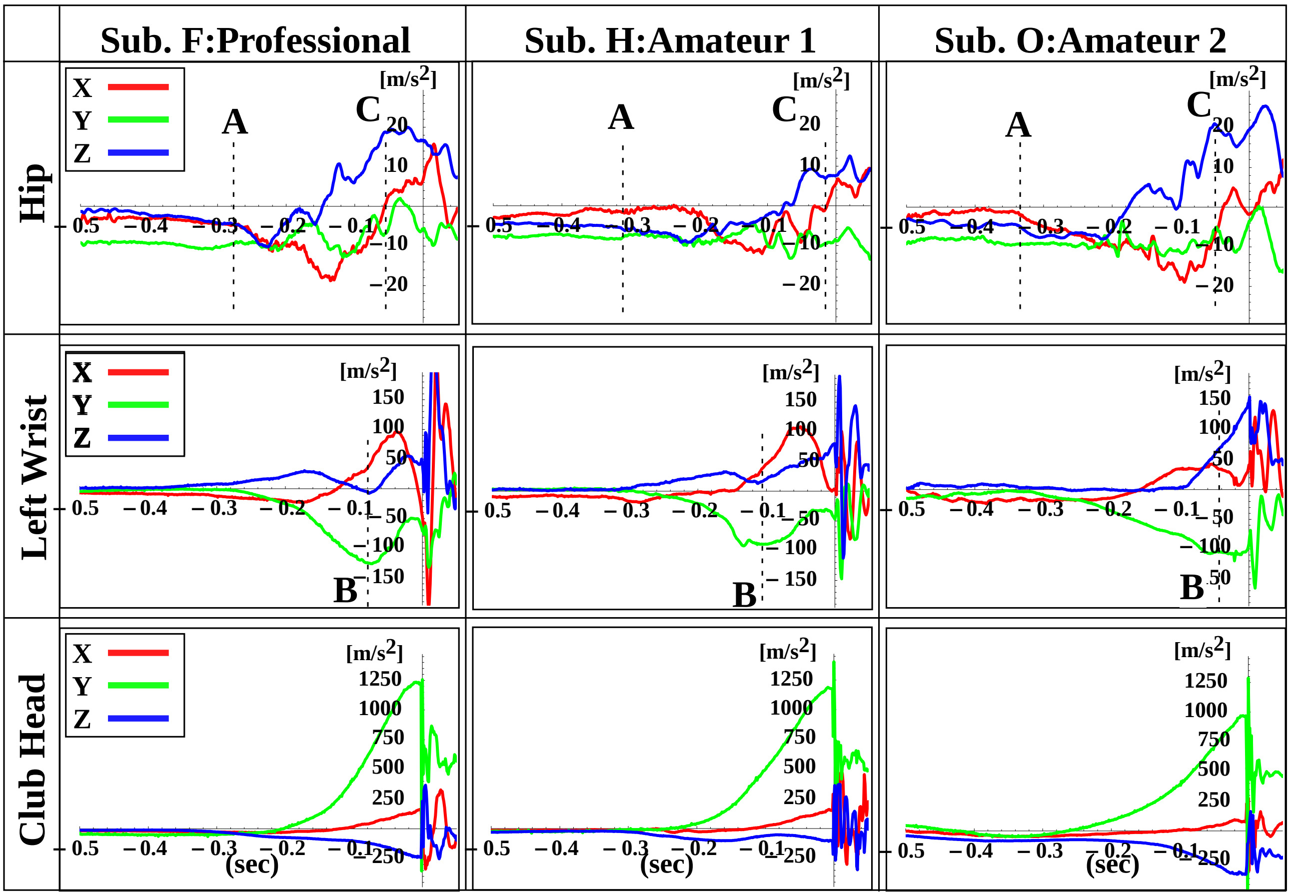


Figure 7: Acceleration time histories of subject's hip, left wrist, and the club head during driver swing.

4. Conclusion

In order to examine the sports skill of the athletes, the author has focused on the kinematical data such as the tri-axial accelerations or angular velocities measured by the inertia sensors attached on their body segments or the equipment. The inertia sensor data, which is measured from the critical location on subject's extremities is sufficient that we can evaluate his sports skill. Then the author has applied those sensor devices to several sports activities, such as the swimming and the golf.

Those inertia sensor devices were in practical use already. For example, pedometer within MEMS acceleromter is manufactured recently. Now, we need an innovative methodology for the interpretation from the sensor time series to the human skill or knowledge.

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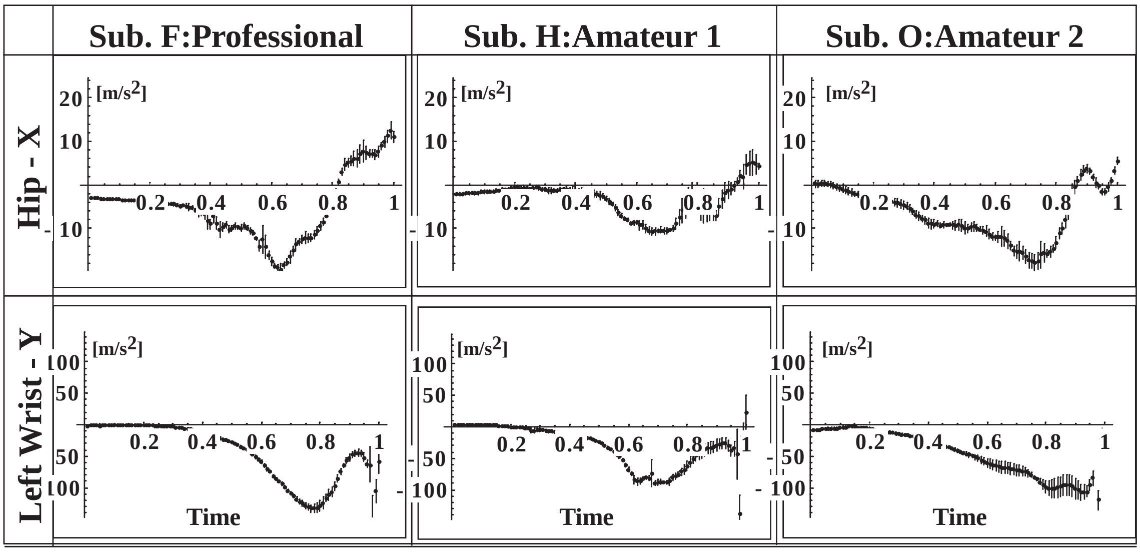

Figure 8: Variability of subject's trunk and left forearm motion. All swing data between –0.4s and the impact instance were normalized to [0,1]. The figure shows the means and the standard deviations of the hip X-axis acceleration and the wrist Y-axis acceleration data at each 0.01 timing during normalized swing phase.

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