

A Distributed Wearable, Wireless Sensor System for Evaluating Professional Baseball Pitchers and Batters

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Abstract

This paper introduces a compact, wireless, wearable system that measures signals indicative of forces, torques and other descriptive and evaluative features that the human body undergoes during bursts of extreme physical activity (such as during athletic performance). Standard approaches leverage high-speed camera systems, which need significant infrastructure and provide limited update rates and dynamic accuracy. This project uses 6 degree-of-freedom inertial measurement units worn on various segments of an athlete's body to directly make these dynamic measurements. A combination of low and high range sensors enables sensitivity for both slow and fast motion, and the addition of a compass helps in tracking joint angles. Data from the battery-powered nodes is acquired using a custom wireless protocol over an RF link and analyzed offline. Several professional pitchers and batters were instrumented with the system and data was gathered over many pitches and swings. We show some biomechanically descriptive parameters extracted from this data, and highlight ongoing work and system improvements.

1. Introduction & Prior Work

Baseball players, especially pitchers, push their bodies to the edge of human capability. Maintaining performance and arm/shoulder health at a professional player level requires close monitoring and interaction with coaching and medical teams, who judge and advise players mainly based on visual observation, allowing players only so many throws per day, etc. Nonetheless, injury can be frequent. In 1973, 50% of pitchers reported shoulder or elbow pain sufficient enough to keep them from throwing [1]. By 1999 that number had grown to over 75% [2]. Andrews [3] reports that from the periods of 1995-1999 and 2000-2004, the need for elbow surgery in professional baseball pitchers increased twofold (for college pitchers fourfold and for high-school pitchers sixfold). Accordingly, there is a need for an easily deployed

system that can allow coaches and sports medicine practitioners to quantitatively measure relevant aspects of player performance. Such data can be leveraged in evaluating and comparing players, better predicting their performance during a game, and identifying and measuring changes in playing technique that either indicates the beginning of an injury or predicts that an injury is probable. If the data from this system can be applied in real time, it could also be useful for interactive therapy during recovery and training.

The most common instrument used to quantitatively measure subjects in sports medicine research clinics is the optical motion tracker [4,5,6]. These systems track the position of reflective markers attached to the body of the subject with an array of high-speed, IR-illuminated video cameras, producing a series of positions that are converted into joint angles to drive a stick-figure animation. These systems typically record in the vicinity of 200 Hz, which can be slow for a detailed analysis of physical performance, as players can exhibit brief extreme bouts of very fast motion. Although higher speed is possible on some models, this tends to require even more detailed setup, calibration, and illumination – constraints that already pose difficulties for rapid implementation and make it impossible to use this system for players in their natural settings – e.g., practicing in the mound, bullpen, or batting cage. Active magnetic motion trackers (such as made by Polhemus and Ascension) [7] can also involve considerable setup and calibration (especially if any ferrous material is nearby), exhibit limited sampling rates if many sensors are used, restrict the sensing area to a small region, and tend to require the pickups to be wired to a large belt-pack. Goniometers, exoskeletons, and mechanical encoders can directly measure joint angle [8,9,10], but require attachment at both ends of a joint, which can be inconvenient and constraining for the players. Similarly, resistive bend sensors integrated into smart garments have been used for motion capture [11], but must be sized to individual subjects and may be subject to speed and reliability limitations when monitoring high-intensity sports gesture.

Due to the very high speeds at which activities like pitching occur, there do not exist any off-the-shelf products that can accurately measure the dynamics (and directly infer the forces and torques) at which the body is moving during peak activity - information that would aid in gaining a better understanding of, for example, the extreme biomechanics of shoulder and elbow. The tracking systems described above must double-differentiate their position measurements in estimating force – a process that is intrinsically noisy. Accelerometers, in contrast, directly measure acceleration, which is proportional to force, providing a much cleaner measurement. Accordingly, designing a player monitoring apparatus around multipoint wireless inertially-measuring sensors promises several advantages, including direct inference of forces and torques, and a simple system of compact, untethered nodes that can be easily applied to the body and used anywhere the team desires, as it doesn't require a structured environment.

Wearable inertial sensors have been used for motion classification and tracking in prior work – for example, in monitoring the activity of people at home, mainly for medical purposes [12,13,14]. SHIMMER [15] is a wearable, wireless health research platform with a 3-axis accelerometer and Bluetooth radio together with a microSD slot for copious on-node data storage. These systems lack sufficient sensor degrees of freedom, sampling rate, and dynamic range for our applications and often employ limited sensing channels to mitigate costs.

Wired [16, 17] and wireless [18,19] systems of nodes employing gyroscopes and/or accelerometers with magnetometers and sometimes even ultrasound have been used for generic motion tracking – these systems lack the dynamic range and sampling rate capabilities needed to make the measurements required for the very fast activities monitored in this study.

Commercial systems that leverage arrays of inertially-sensing nodes have recently been appearing – these are primarily aimed at the healthcare, physical therapy and biomotion measurement markets – for example, the IDEEA LifeGait System from Mini-Sun uses wired accelerometers distributed on the body to determine parameters of gait and motion [20].

Some other research groups have applied wireless inertial sensor nodes to sports. Recent projects, for example, have explored interactive golfing practice [21, 22], and an early study [23] took an initial look at pitching analysis with wired accelerometers. Wireless, limb-mounted inertial nodes have also been used for exercise coaching [24] and mobile interactive entertainment driven by exercise [25]. Again, the dynamic range and sampling rate for these systems are far below our goals for this project.

Our work has evolved from a lineage of projects at the MIT Media Lab. The Lab's initial foray into baseball was an interactive batting analyzer by Gerasimov called "Swings That Think." [26]. Here, a gyro wired into a bat informed a classifier that would output a simple verbal assessment after each swing. The project discussed in this paper came from systems that our research group designed for interactive dance. Initially focusing on a highly instrumented pair of wireless sensor shoes for a solo dancer [27], this work evolved into an array of compact wireless IMU's that each conversed with a common basestation using a fast 1 Mb/s radio and a simple TDMA protocol. Termed "Senseble" [28], this setup was designed for a real-time instrumented dance ensemble, with each dancer wearing nodes at the wrists, ankles, and perhaps head. Fast data fusion at the base station and host computer produced a finite set of parameters that a composer or choreographer could author interactive content upon. Our first proof-of-concept baseball nodes adopted the Senseble system, adapting it to instrument pitchers by replacing the 300°/s gyros and 5-G accelerometers with gyros that range up to 11,000°/s and accelerometers that work up to 120 G's in order to sample peak pitcher dynamics [28, 29, 30]. The system would acquire 5.6 seconds of 1 kHz sampled data in onboard flash memory after the user indicated that sampling should begin via a synchronization and activation message sent wirelessly from the base station – data could subsequently be read off all nodes via our custom TDMA downlink. Biomechanical results from these tests, made with players at Red Sox spring training in 2006, were summarized in [30].

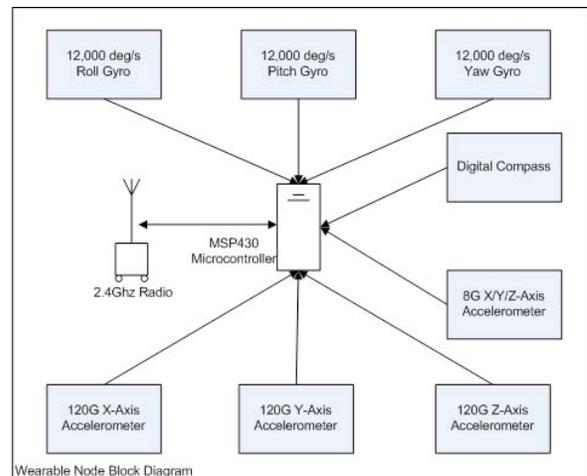


Figure 1: Block Diagram of a SportSemble Node

2. The SportsSemble System

We subsequently evolved our Senseable system into a node dedicated to wearable monitoring of athletic gesture, specifically for measuring baseball players' pitching and batting. A block diagram of our resulting SportSemble node is given in Figure 1, a photograph of an actual node is shown in Figure 2, and more details on the design of this system can be found in [31]. Each node measures about 2.2 inches by 2.0 inches and weighs 44 grams with battery and attachment bracelet. The battery is a 145mAh lithium polymer rechargeable that was seen to continuously power a node for up to 3 hours of use. Two orthogonal daughter cards (in addition to the main board) house high-range gyros and accelerometers in order to span all 6 inertial axes. The embedded radio (a 2.4 GHz Nordic nRF2401a, which has a maximum bandwidth of 1 Mbps and an output power of +4dBm) is mounted on a daughter card to isolate the RF electronics from the main board and enable an easy upgrade of the RF hardware. The embedded radio interfaces to the microcontroller directly via a Serial Peripheral Interface (SPI). Each node has a unique address, and radio communication to the base station exploits a custom frequency-hopping protocol described below. A node consisting only of microcontroller, radio, and USB interface is connected to a laptop and works as the base station for all nodes.



Figure 2: Photograph of working SportSemble node

Each of the wearable battery-powered nodes has three single-axis $\pm 120G$ ADXL193 accelerometers, three single-axis ADRX300 gyroscopes, a 3-axis $\pm 8G$ LIS302DL accelerometer and a HMC6343 digital compass. The bi-range accelerometers let us record slow motion with the low-G device, and fast motion with the high-G units, thus providing high resolution across the entire gesture. The gyros, which normally

saturate at $\pm 300^\circ/s$ were specially biased and strapped to respond up to $12,000^\circ/s$, as described in [32] and originally implemented in our earlier trials [28].

The output from the 3 analog High-G accelerometers (which have a 400 Hz bandwidth) and gyroscopes (which have a 2 kHz bandwidth) are digitized into 12 bits by the microcontroller at a 1 kHz rate after passing through first-order analog filters to suppress noise and/or basic aliasing. The additional sensors (low-G accelerometer and compass) talk via a common digital I²C interface. The low-G accelerometer is sampled at 100 Hz and the compass at 10 Hz – adequate for measuring the slow motions that they were intended for. The onboard MSP430 microcontroller used with the SportSemble had 116 kB of onboard flash memory, enabling each node to store about 11 seconds of data sampled at 1 kHz.

3. Communications and Embedded Code

At a high level, the firmware running on the nodes is simple: wait for a basestation beacon via the radio and react based on the command encoded in the beacon's packet. Reacting can be one of 4 things: do nothing, erase flash, begin sampling data to onboard flash, or return a sample to the base station. The firmware for the base station also has 3 basic functions; simple beaconing at 200Hz (these packets contains no data, but simply ensure that all nodes keep synchronization), receiving commands via the USB, transmitting them to the network of nodes when appropriate, and returning the results of commands via USB to the host laptop.

When a data collection run is initiated, a command is broadcast from the base station that erases the contents of Flash memory on all nodes. The nodes then wait until they receive a "Sample2Flash" command, at which point they sample all sensor signals as described earlier and write the results to Flash memory, proceeding until the Flash is full. A full-color LED onboard each node changes color to designate the node's state – initialized and erased, sampling, or full. Once sampling is completed, the data is read from each node byte-by-byte, with the base station requesting sequential packets from the nodes. If a received data byte fails a CRC test, it is re-requested – this insures that all node data arrives at the base station and host computer intact. Data can be wirelessly read from all five wearable nodes used in our tests within 35 seconds in good RF conditions. To mitigate problems with RF interference, a simple frequency-hopping scheme is employed on readout that sequentially jumps to an adjacent channel after the node transmits the desired data record. A control

packet, which can be sent when the nodes are initialized, sets the base RF channel used and the hop count – this way the channels can be selected to avoid noisy bands in different environments where the system is deployed. More details on our communication scheme can be found in [31].



Figure 3: Setup with reference camera tracking system (top), pitcher w. targets (left) & reconstruction (right)

4. System Deployment & Test

Before the nodes were used for field measurements, all inertial sensors were tested and calibrated on a variable speed rotating platform, as described in [31]. To evaluate this system, we compared the IMU array to an optical tracking system. A series of professional baseball players was used for this experiment. After IRB approval and informed consent and under the direction of the subjects' coaches, a series of 4 professional baseball pitchers underwent simultaneous biomechanical testing utilizing both the camera-based motion tracking system and our newly-developed IMU array. Strobe signals from the optical tracker were also sampled by our base station in order to synchronize and align the data taken by the optical and wearable systems [31].

The optical motion analysis system (from XOS Technologies), employing “high-speed” cameras operating at 180 Hz, allowed positional tracking of each pitch. A series of 10 motion analysis cameras

were set-up on a regulation-sized pitching mound (Figure 3). Subjects were fit with both passive reflecting targets for the camera-based motion analysis and a 5 segment wireless IMU array. Inertial measurement units were carefully affixed to the chest, upper arm, forearm, wrist, and waist (Figure 4) with standard body tape, such as sports pre-wrap [31].



Figure 4: Nodes initially applied to player (left) and fully taped up during data taking (right)

When the player was ready to record a pitch, swing or baseline calibration, the operator entered a laptop command that instructed the base station to transmit a “Sample2Flash” code, putting all nodes into acquisition mode. After the 11 seconds of sampling elapsed, another laptop command was entered to dump all data back to the base station and laptop (the player could continue throwing or swinging during this period, although this data wasn't recorded). Once all data was transferred, all flash was erased, and the above process was repeated to record more data.

Each pitcher threw ten recorded fastballs using a regulation baseball off a regulation pitcher's mound. Using positional data from the camera-based tracking system, real-time 3D cartoon reconstructions of each pitch were made (Fig. 3, right). The acceleration phase of the pitching cycle was isolated, and kinematic parameters of the shoulder were calculated from simultaneous recordings of position, acceleration, and velocity by the two systems [31]. Maximum acceleration and velocities were compared at the wrist, shoulder, and hand. A commercial radar gun monitored the speed of each pitch, which was manually logged.

The system was then utilized to evaluate a series of five professional baseball batters. Players were fit with a 5-segment wireless IMU array. Inertial measurement units were affixed to the chest, upper arm, forearm, and to the bat itself. Players were then asked to perform 5 free swings. The IMU node was removed from the bat and affixed to the waist. Players then hit a ball off a tee for five additional swings. Relative speeds of the

bat, hands, forearm, and chest were calculated. Ball/bat Impact times were calculated and hand speed at impact determined.

In the results presented below, an assembly error in the first order front-end filter limited the analog bandwidth of the IMU signals to ~50 Hz. We examined the impact of this in our data in two ways – by compensating the low passed data offline by a matched high pass filter (rolling off noise at high frequency), and also by taking representative data with the hardware filters working properly. Although there may be subtle fine structure in the higher bandwidth data, the peak values, averages, and timings derived from the low-passed data presented here look to differ from full bandwidth results by under 3%, hence any distortion in these results look to be marginal.

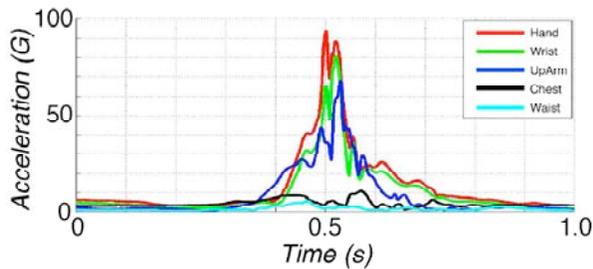


Figure 5: IMU-Measured Multipoint Accelerations during a pitching cycle

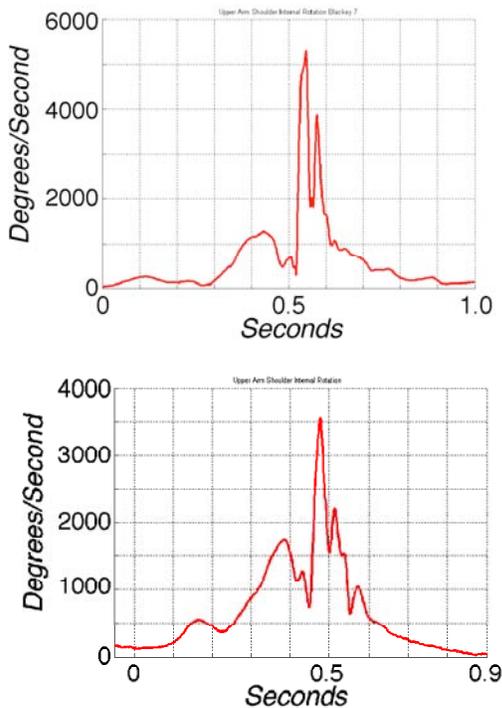


Figure 6: IMU-inferred shoulder internal rotation over time during the pitch cycle for two different players

Location	Average Peak g-Forces
Hand	90
Wrist	80
Forearm	70
Chest	10
Waist	8

Table 1: Average G-Forces in Pitch Acceleration

5. Results

5.1. Pitching.

A rapid rise in elbow extension velocity and humeral internal rotation was recorded in both the optical system and with the wearable IMUs. The acceleration phase of the pitching cycle (Figure 5) lasts from peak external rotation of the shoulder to the time of ball release and was identified as a sudden peak in the acceleration forces in the hand and upper extremity. The acceleration components plotted here point along the centripetal axis (directed down the arm), hence are unipolar. In our series, the average acceleration phase lasted 0.022 seconds. The high-speed motion-tracking camera system was able to capture four data points during this phase of the pitching cycle. The IMU array captured 30 data points during this same period.

Resultant g-forces were calculated from the IMUs. Average g-forces at the hand are depicted in Table 1. Peak hand forces averaged 90 g's in the acceleration phase of the pitch and were appropriately captured by the accelerometer chosen. As expected, g-forces at the chest and waist were considerably slower at 10 and 8 respectively.

Shoulder internal rotation values were estimated from the IMU measurements as detailed in [31]. Graphs of shoulder internal rotation velocities demonstrated a similar pattern, as throwing mechanics are very similar at this level of participation. Each graph, however, demonstrated a unique signature describing the individual's shoulder mechanics. (Figure 6)

Peak shoulder internal rotation velocity was identified for each pitch – averages and standard deviations were calculated between all throws made by each player (Table 2). There was no statistical difference between average shoulder internal rotation velocity measured from the optical system compared to the IMU array. The average standard deviation of the IMU array was about 6% whereas the average standard deviation of the optical system was 15%.

		IMU Array		Optical	
Pitcher		Average (deg/sec)	STD Dev	Average (deg/sec)	STD Dev
Subject 1	74 mph	3023	680	4586	1612
Subject 2	81 mph	4586	104	4976	362
Subject 3	81 mph	4931	318	5574	395
Subject 4	74 mph	4991	138	5218	515
TOTALS		4431 ± 291		5080 ± 791	

Table 2: Average shoulder internal rotation velocity (deg/sec) for each subject. Average pitch velocity (from radar) is listed in the first column on the left.

5.2. Batting.

A similar rapid rise in g-forces and angular velocities were seen during batting swings (Figures 7 & 8). Due to the very high shock transmitted through the bat at impact, the node was removed from the bat during swings that used a tee to avoid IMU damage.

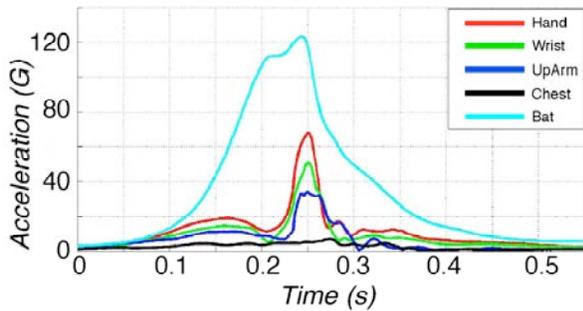


Figure 7. IMU-Measured G-forces during a representative swing of a baseball bat (free swing)

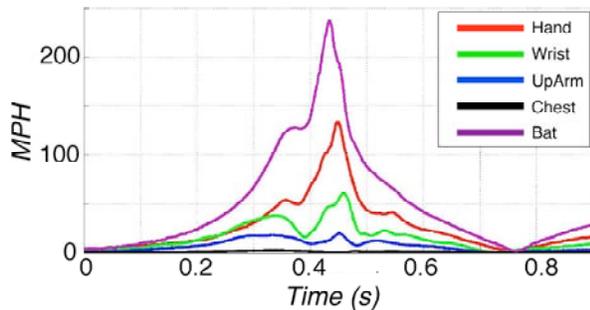


Figure 8. IMU-Derived swing speeds from the bat's sweet spot during a representative swing (free swing)

Impact could be detected from the hand node, as seen by a disruption of the bat speed curve (Figure 9). Relative body segment velocities & timing differences were determined for each swing (Figure 10) [31].

Average bat speed during a free swing was 227.5 MPH. Average standard deviation was 9.7 MPH. Average hand speed at impact was 74.5 MPH.

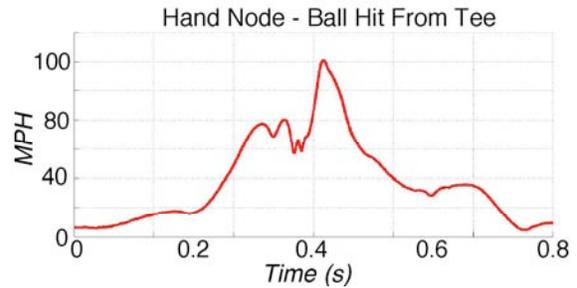
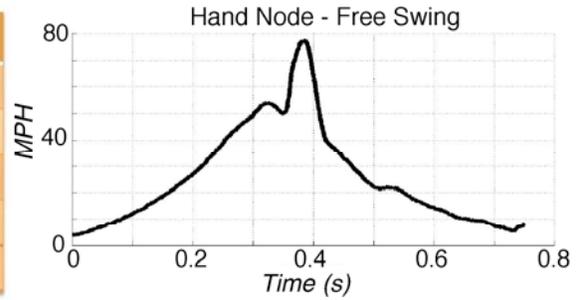


Figure 9: Hand node resultant velocities during a free swing (top) and after hitting a ball from the tee (bottom).

Approximate bat speed can be predicted from hand speed by knowing the length of the bat and its angular velocity. This calculation was used to predict bat speed at the time of impact when a node was removed from the bat itself (nodes cannot withstand the force of impact if left on the bat). For each of the 5 players, this calculation was verified using the free swing data (where a node was left on the bat for each swing). Average error in this calculation was $4.8 \pm 0.5\%$.

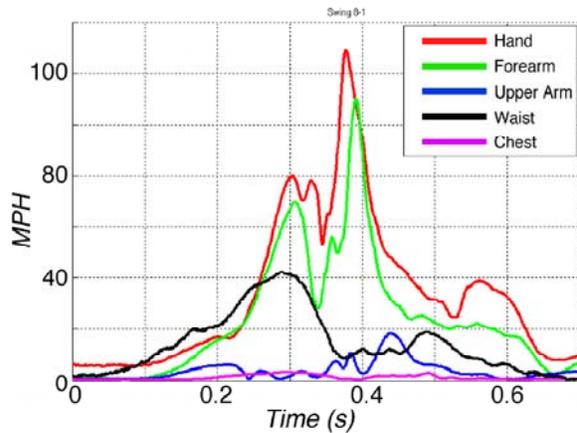


Figure 10: Graphs of body segments over time. Impact is seen at 920ms. Hand speed at ball impact was 74MPH

7. Conclusions and Future Work

We have presented results from, to the best of our knowledge, the first application of an array of wireless inertial measurement units to measuring dynamic and

kinematic properties of pitchers and batters in action. Our system provides a self-contained wireless data acquisition studio, where complex fixed infrastructure, such as needed by standard optical tracking systems, is not needed. We have seen indications that the calculation of forces and torques with the inertial system (measurements of special interest in player evaluation and injury prediction) seem to be much cleaner than those inferred from the optical system. We have performed some crosschecks to validate that the data from the inertial system is compatible with that produced from the optical tracker – validating joint angle estimates across both systems awaits further analysis, however, as described below. Our system was seen to sample 7-8 times more quickly than the optical tracker used in these tests, and provides much more granularity in the measurements, which should be extremely useful when analyzing the extremes of athletic motion (e.g., the critical acceleration peak for pitching was only sampled at 4 points by the optical system). Although the error in the pre-sampling analog filters prevented our system from measuring high-frequency data in these tests, preliminary testing has indicated that the results presented here are still valid. This problem has now been fixed, and upcoming tests will exhibit full bandwidth, which will enable the analysis of fine structure in pitching dynamics.

Persistent difficulties with the embedded software before our spring training deadline prevented the dataset presented here from also including information from the magnetometer (compass) and low-G accelerometer – also, although our high-rate gyro measurements produced good data, lower rate phenomena (such as during pitch windup, etc.) had insufficient signal-to-noise to enable a useful integration across a pitch.

Accordingly, we have incorporated lessons learned here into a new SportSemble node design (Figure 11). The basic system is identical to the one used in the study presented here, except that we have also added 3 axes of low-rate gyros to enable good signal-noise performance for low angular rate activity, enabling us to have very wide dynamic range in both accelerometer and gyro subsystems (ultimately, one would like a log response in these components for sports research, but log inertial sensors appear to be thusfar unavailable). We have also upgraded the embedded processor to an AVR32 – this device proved to be much more capable than the MSP430, making the embedded software design easier, hence resolving the problems that we encountered in the tests presented here with reading the low-G accelerometer and the magnetometer. Finally, our new design supports a micro-SD slot, allowing us to install copious memory. This has revolutionized our testing – now we log all data right into the slotted flash

chip. Accordingly, we use the radio only for synchronization and transmission of basic commands – bulk data is RF-transferred from the nodes only on occasion for spot checking. After a day of running, the SD memory cards are manually removed from the nodes' slots and their contents are copied to a PC before they are erased for subsequent tests. This allows us to operate much more quickly out on the field, and record every pitch and hit that the players generate. We have recently gone to spring training to take new data with this design, and analysis results are forthcoming.

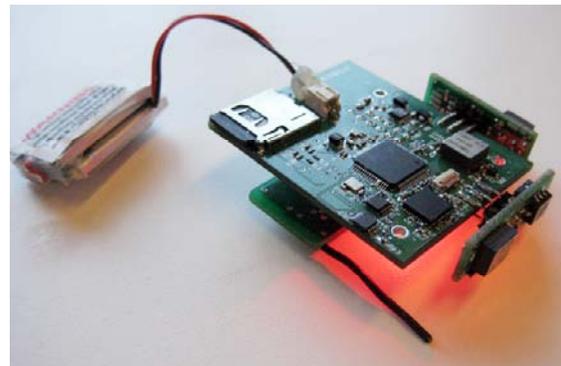


Figure 11: New SportSemble Node with AVR32, 12 inertial sensors for 6-axes of dual range, & SD-slotted memory.

Our new tests will provide us the capability of fusing data from all 12 inertial sensors with the 3 magnetometer values to estimate joint angles – this should allow us to track body position across very high dynamic range.

Although the RF datalink worked well in our lab, when moving to an outdoor location without nearby walls to reflect the nodes' signals, we encountered substantial problems in signal absorption by the body – when the player was facing away from the base station, our bit error rate could become very high. Although this is much less of a problem with the new system, which doesn't rely on the RF channel to download data, putting a power amplifier on the base station's transmitter or moving to a sub 900 MHz carrier that exhibits less corporal absorption would allow commands to be very reliably received.

Further development with this system will work on shrinking the nodes, improving the user-interface software, and making it much easier to apply and remove in the field, allowing coaches to use it as a standard tool. Systems like ours promise to soon enable wide-ranging practical and clinical applications for athletes, including injury prevention (youth pitching), aiding in conditioning/training, and improving post-operative rehabilitation

8. Acknowledgements

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