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Analysis of Athlete Training Data Based on Intelligent Sensing Technology for Sports Clothing

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Article

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ABSTRACT

This paper details the design, fabrication, and performance validation of a functional electronic textile (e-textile) engineered for biomechanical monitoring. A warp-knitted nylon-elastane compression garment serves as the substrate, a material choice critical for ensuring a stable and conformal skin-sensor interface. The sensing functionalities are directly integrated into the fabric structure. Electromyography (EMG) electrodes were fabricated by screen-printing a stretchable, silver-based conductive ink to monitor key muscle groups, while flexible piezoresistive sensors were incorporated to function as a textile-based goniometer for kinematic tracking at the knee joint. The performance of this integrated smart garment system was rigorously evaluated in a study with 12 runners performing a run to exhaustion. The system's data output demonstrated high fidelity when validated against laboratory-grade motion capture systems (R2 > 0.95). Crucially, the results confirmed the textile's ability to reliably measure the physiological and biomechanical markers concurrent with the onset of neuromuscular fatigue, evidenced by a significant decrease in EMG median frequency (p < 0.01) and concurrent biomechanical alterations, where changes in Ground Contact Time were highly significant (p < 0.01) and Knee Flexion Range of Motion was significant (p < 0.05). This study validates the integrated textile system as a high-performance and reliable platform, demonstrating the viability of using engineered fabric structures for complex, dynamic human movement analysis.

KEYWORDS

e-textiles, textile sensors, conductive yarns, functional fabrics, running biomechanics

INTRODUCTION

The pursuit of peak athletic performance is a multifactorial endeavor, heavily reliant on structured training, strategic recovery, and the avoidance of injury [1,2]. Central to this process is the objective assessment of an athlete's physiological state and biomechanical efficiency [3]. Historically, such assessments have been confined to specialized sports science laboratories equipped with sophisticated instrumentation like 3D motion capture systems (e.g., Vicon, Optotrak) and stationary electromyography (EMG) units [4-6]. While these technologies provide gold-standard data, their application is inherently limited. The high cost, requirement for technical expertise, and controlled, artificial environment prevent their use for continuous, longitudinal monitoring during daily training activities [7,8]. This gap between laboratory analysis and real-world application has created a significant demand for portable, unobtrusive, and valid monitoring solutions that can provide ecologically valid data.

The emergence of wearable technology has begun to address this need, with devices such as GPS watches and heart rate monitors becoming ubiquitous in both amateur and professional sports [9,10]. However, these first-generation wearables primarily capture systemic physiological data or gross movement metrics, often lacking the specificity to detail the nuanced biomechanical and neuromuscular changes that are precursors to performance decrements and injury [11,12]. For instance, while heart rate can indicate overall cardiovascular load, it provides little insight into localized muscle fatigue or compensatory changes in movement patterns. A more sophisticated approach is required to capture the intricate interplay between the neural, muscular, and skeletal systems during dynamic activities.

It is in this context that smart textiles, or e-textiles, have emerged as a revolutionary platform for next-generation wearable sensing. By integrating electronic functionalities directly into the fabric structure, smart textiles offer a paradigm shift away from rigid, strapped-on devices. Sensors can be screen-printed, embroidered, or woven into the garment, resulting in a system that is conformal, comfortable, and moves seamlessly with the athlete's body [13]. This intimate skin-sensor interface minimizes motion artifacts and allows for the precise targeting of specific anatomical locations [14]. For textile science and engineering, this represents a significant frontier, focusing on the development of conductive yarns, flexible substrates, and durable encapsulation methods that can withstand the mechanical and chemical stresses of athletic use, including stretching, abrasion, and repeated laundering.

Despite promising advancements in textile sensor technology, much of the existing research has focused on single-modality sensing, such as monitoring respiration via strain-sensitive fabrics or cardiac rhythm via textile electrodes. While valuable, these isolated data streams fail to capture the holistic nature of athletic performance. Fatigue, for example, is not merely a cardiovascular or muscular phenomenon; it is a complex state involving central and peripheral mechanisms that manifest as concurrent changes in muscle activation strategies and kinematic patterns [15,16]. Therefore, a multi-modal sensing approach, integrated within a single garment, is necessary to provide a comprehensive analysis. This study addresses this specific research gap by focusing on a small but critical area: the detection of running-induced fatigue through the simultaneous analysis of neuromuscular activity and joint kinematics. The primary objective was to design, fabricate, and validate a smart textile system capable of quantifying the distinct biomechanical and myoelectric signatures of fatigue in runners, thereby demonstrating the viability of this technology for advanced, field-based athletic monitoring.

MATERIALS AND METHODS

Smart Garment System Design and Fabrication

The core of the data acquisition system was a pair of custom-designed smart compression tights. The base garment was fabricated from a high-performance, warp-knitted blend of nylon (80%) and elastane (20%), selected for its high stretch recovery, moisture-wicking properties, and ability to provide consistent compression. This ensures a stable and intimate contact between the integrated sensors and the skin, which is paramount for minimizing motion artifacts and ensuring signal quality.

The neuromuscular sensing modality was achieved through textile-based EMG electrodes. These electrodes were specifically designed to monitor the activity of two key muscles in the running gait cycle: the rectus femoris (RF), a primary knee extensor, and the biceps femoris (BF), a primary knee flexor and hip extensor. The electrodes were fabricated by screen-printing a stretchable conductive ink, comprising a silver/silver chloride (Ag/AgCl) composition within a polyurethane binder, directly onto the inner surface of the fabric. Each EMG sensor site consisted of two active electrodes (20 mm diameter) with an inter-electrode distance of 25 mm, and a common reference electrode placed over the distal iliotibial band (ITB), proximal to the lateral femoral epicondyle. While clinical guidelines (e.g., SENIAM) recommend bony prominences for reference electrodes, preliminary testing indicated that the significant skin deformation over these rigid structures during dynamic running caused severe motion artifacts. Therefore, the distal ITB was selected to prioritize electrode-skin contact stability under the garment's compression. This design choice aims to minimize motion-induced noise, accepting a theoretical trade-off regarding far-field crosstalk which is mitigated by the distance from the active muscle bellies. The locations were determined based on SENIAM (Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles) guidelines to ensure accurate and repeatable measurements.

Kinematic sensing was integrated to measure the knee flexion angle. This was accomplished using flexible piezoresistive sensors. Each sensor consisted of a thin strip of conductive polymer whose electrical resistance changes predictably with mechanical deformation (bending). The sensors were encapsulated in a thin, flexible silicone elastomer to provide a hermetic seal against ionic sweat saturation. They were then securely stitched onto the exterior of the garment. This exterior integration was a deliberate design choice to create a 'triple-layer' stability mechanism: (1) Ionic Isolation via silicone; (2) Thermal Buffering, utilizing the textile substrate to insulate the sensor from direct skin heating; and (3) Mechanical Locking, where the garment's compressive pre-tension dominates the sensor-fabric interface, minimizing baseline drift caused by hygroscopic fiber swelling. The sensor was pre-calibrated to establish a precise relationship between its resistance change and the corresponding flexion/extension angle in degrees. This static calibration was performed for each participant prior to the run by recording the sensor's resistance values at 0°, 45°, and 90° of knee flexion (as measured by a standard goniometer) to establish a linear calibration function. While piezoresistive materials inherently exhibit strain-rate dependency and hysteresis, the running gait cycle frequency in this study (≈ 1.2 - 1.5 Hz) falls within the sensor's bandwidth of quasi-linear response. Therefore, a static calibration model was deemed a sufficient approximation for capturing the gross kinematic patterns (Range of Motion), acknowledging that minor phase lags may exist which do not impact the peak-to-peak amplitude measurement.

A lightweight, detachable Data Acquisition Unit (DAU) served as the electronic hub of the system. The DAU, weighing 35 grams and housed in a 3D-printed ABS enclosure, was mounted at the rear waistband of the tights. It contained a high-resolution 24-bit analog-to-digital converter (ADC) for the EMG signals, a separate 16-bit ADC for the piezoresistive sensors, a low-power microcontroller (ARM Cortex-M4), and a Bluetooth Low Energy (BLE) 5.0 module for wireless data transmission to a host device (e.g., a smartphone or laptop). The entire system was powered by a rechargeable 3.7V, 250mAh lithium-polymer battery. Conductive silver-coated nylon threads were used to create soft, flexible interconnects between the sensors and the DAU, which were routed along the garment's seams to maximize comfort and durability. A schematic of the complete system is shown in Figure 1.

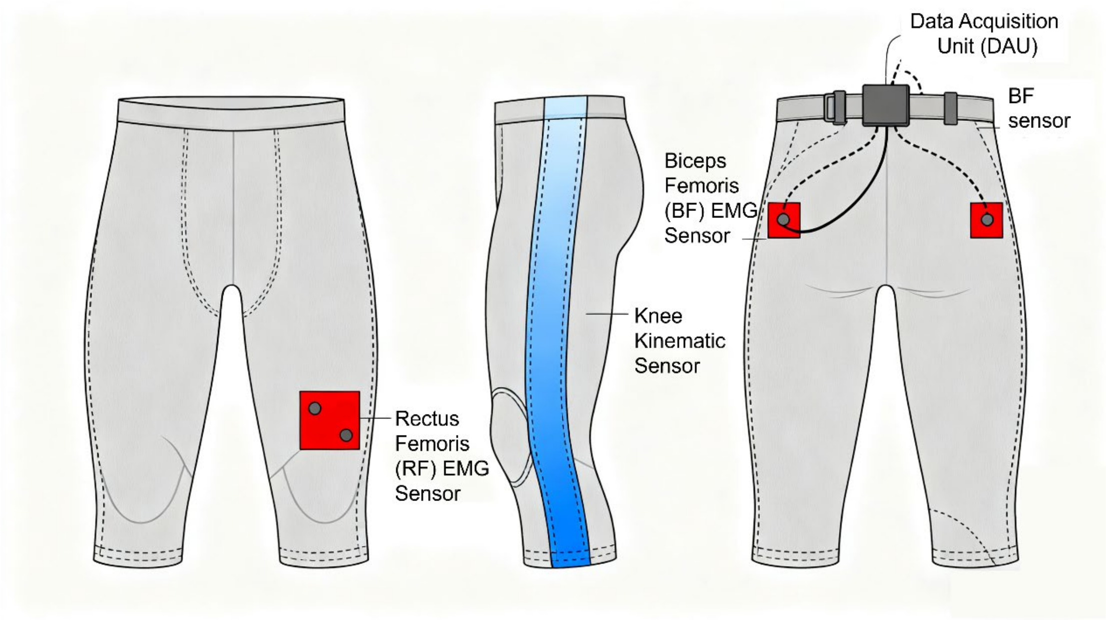

Figure 1. A schematic diagram of the smart compression tights, illustrating the placement of the EMG electrodes on the rectus femoris (RF) and biceps femoris (BF), the piezoresistive kinematic sensor at the knee joint, and the location of the detachable Data Acquisition Unit (DAU). The diagram depicts the two-electrode configuration at the muscle sites, with the common reference implied at the lateral thigh

Experimental Protocol and Participants

Twelve healthy male recreational runners were recruited for this study (age: 26.4±3.1 years; height: 178.5±5.2 cm; body mass: 72.3±6.8 kg; weekly running distance: 35±10 km). All participants provided written informed consent prior to participation. The study was approved by the Institutional Ethics Committee and conducted in accordance with the Declaration of Helsinki.

Each participant completed one experimental session in a controlled laboratory environment. Upon arrival, participants were fitted with the smart textile system. To validate the sensor data, they were concurrently instrumented with a gold-standard 3D motion capture system (Vicon, UK) with reflective markers placed on key anatomical landmarks of the lower limb, and conventional pre-gelled Ag/AgCl surface electrodes (Ambu BlueSensor N) placed adjacent to the textile EMG sensors.

Prior to the treadmill running protocol, a Maximal Voluntary Contraction (MVC) test was performed to establish a reference for EMG normalization. Participants were seated on an isokinetic dynamometer (or a custom-built rigid chair) with the hip and knee joint angles standardized. They performed three 5-second maximal isometric contractions for the rectus femoris (knee extension) and biceps femoris (knee flexion) with a 2-minute rest interval between trials. The highest RMS value obtained from these trials was recorded as the MVC reference (RMS_{MVC}). Prior to the main experimental visit (at least 48 hours in advance), participants completed a preliminary incremental treadmill test to determine their maximal oxygen uptake (VO_2max). Gas exchange was measured breath-by-breath using a calibrated metabolic cart (Quark CPET, COSMED, Rome, Italy). The protocol commenced at a speed of 8 km/h (1% gradient) with speed increments of 1 km/h every minute until volitional exhaustion. VO_2 max was defined as the highest 30-s average oxygen uptake value meeting at least two of the following criteria: a plateau in VO_2 despite increasing workload, a respiratory exchange ratio (RER) > 1.10, or a heart rate within 10 beats/min of the age-predicted maximum. The experimental protocol consisted of a continuous run on a motorized treadmill at a constant speed, corresponding to 80% of their previously determined maximal oxygen uptake (VO_2 max).

The protocol began with a 5-minute warm-up at a self-selected comfortable pace. Following the warm-up, the treadmill speed was increased to the target intensity. Participants were instructed to run for as long as possible until they reached the point of voluntary exhaustion, defined as the inability to maintain the required pace despite verbal encouragement. Rating of Perceived Exertion (RPE) on the Borg 6-20 scale was recorded every five minutes. While systemic metabolic markers (e.g., blood lactate) were not recorded, the occurrence of *neuromuscular* fatigue—as distinct from solely cardiorespiratory exhaustion—was confirmed by the significant spectral shift (decrease in Median Frequency) observed in the EMG analysis.

Data from all systems (smart textile, Vicon, and conventional EMG) were continuously recorded and synchronized. For subsequent analysis, two distinct 60-second epochs of data were extracted:

- Non-Fatigued (NF) State: To eliminate the confounding variable of running speed, data were extracted from the 3rd minute after reaching the target intensity (i.e., total time 08:00 to 09:00), ensuring participants had reached a physiological steady state.
- Fatigued (F) State: The final 60 seconds of data immediately preceding the participant's voluntary termination of the test.

Data Processing and Statistical Analysis

The data acquisition sampling rates were configured to 2000 Hz for the EMG signals and 100 Hz for the kinematic data, ensuring sufficient temporal resolution to capture rapid neuromuscular events and dynamic joint angles. All data processing and analysis were performed using custom scripts in MATLAB (R2023a, The MathWorks, Inc.). To ensure temporal synchronization between the wireless textile DAU and the laboratory systems (Vicon and conventional EMG), a mechanical sync event (three rapid foot stomps) was performed by the participant at the start of each recording. These events generated distinct signal spikes across all modalities (ground reaction force/kinematics in Vicon, and acceleration/EMG artifacts in the textile system), which were used to manually align the time-series data during post-processing. The raw EMG signals from both the textile and conventional electrodes were first band-pass filtered between 20 Hz and 450 Hz using a 4th-order Butterworth filter to remove motion artifacts and high-frequency noise. The kinematic data from the piezoresistive sensor and the Vicon system were low-pass filtered at 10 Hz.

For the EMG analysis, two primary features were extracted from the filtered signal for each muscle (RF and BF). First, the signal amplitude was quantified by calculating the Root Mean Square (RMS) value over the 60-second epoch. To facilitate comparison between states, these RMS values were subsequently normalized. To enable comparison of muscle activation levels, the RMS values from the running epochs were normalized to the Maximum Voluntary Contraction (MVC) values obtained prior to the run. The normalized RMS amplitudes are expressed as a percentage of MVC (%MVC), calculated as: $RMS_{\%MVC} = (\frac{RMS_{running}}{RMS_{mvc}}) \times 100$ Second, to assess neuromuscular fatigue, a spectral analysis was performed. The power spectral density (PSD) of the signal was estimated using Welch's method, and the Median Frequency (MDF) was calculated. A decrease in MDF is a well-established myoelectric manifestation of localized muscle fatigue.

For the kinematic analysis, individual gait cycles were identified from the knee angle data. From this, two key biomechanical parameters were calculated: the knee flexion range of motion (RoM) during the swing phase (derived from the textile's kinematic sensor) and the ground contact time (GCT). As the textile system did not include foot sensors, GCT was determined from the Vicon foot marker data to serve as a key contextual marker for changes in gait strategy, measured concurrently with the textile sensor data.

The validation of the smart textile sensors was performed by comparing their output against the gold-standard systems during the NF state. The knee angle measured by the piezoresistive sensor was compared to the angle calculated by the Vicon system using Pearson's correlation coefficient (R) and the coefficient of determination (R^2). The EMG signals from the textile and conventional electrodes were compared by correlating their RMS values.

To test the primary hypothesis, the extracted features (MDF, RMS, Knee Flexion RoM, GCT) were compared between the Non-Fatigued (NF) and Fatigued (F) states. A paired-samples t-test was used for each variable to determine if the changes induced by fatigue were statistically significant. The level of significance was set at $p < 0.05$.

RESULTS

Sensor System Validation

The validation analysis confirmed the high fidelity of the smart textile sensing system. The knee flexion angle measured by the integrated piezoresistive sensor showed excellent agreement with the data from the Vicon motion capture system. A representative plot for one participant is shown in Figure 2, illustrating the close tracking of the two signals across multiple gait cycles. The knee flexion angle measured by the textile sensor showed an excellent correlation with the Vicon system (Pearson's r = 0.98 ± 0.02), corresponding to a coefficient of determination $R^2 > 0.95$. Values are reported as mean ± standard deviation across participants (n = 12). To explicitly evaluate the dynamic performance of the static calibration model, absolute error metrics were calculated. A Root Mean Square Error (RMSE) of 4.1±1.3° was observed across the gait cycle. Furthermore, a systematic phase lag of 32±8 ms was detected in the textile signal relative to the Vicon system, attributed to the inherent hysteresis of the piezoresistive polymer. However, this lag was consistent across trials and did not significantly affect the peak-to-peak Range of Motion (RoM) calculation ($p > 0.05$), validating the sensor's utility for extracting gait amplitude metrics despite minor temporal delays. Crucially, to assess for potential signal drift or hysteresis from prolonged use, mechanical strain, and sweat, this validation analysis was repeated on the 60-second epoch from the Fatigued (F) state. The correlation between the textile sensor and the Vicon system remained exceptionally high (e.g., R=0.97±0.03; $R^2 > 0.94$), with no significant mean offset drift observed. This confirms the sensor's stability throughout the protocol and validates that the detected ~ 8-degree change in RoM was a physiological change, not an artifact of sensor drift. Similarly, the RMS values of the EMG signals from the textile electrodes were highly correlated with those from the adjacent conventional gel electrodes (R=0.96±0.03), confirming their efficacy in capturing muscle activation amplitude. Crucially, to assess the validity of the distal ITB reference placement, the signal-to-noise ratio (SNR) and baseline noise levels (resting state) were compared. The baseline RMS noise for the textile system (1.2±0.4 %MVC) showed no significant difference compared to the conventional gold-standard configuration (1.1±0.3%MVC; p=0.42). This confirms that placing the reference electrode on the distal ITB did not introduce significant far-field crosstalk or motion artifacts, supporting its suitability for the dynamic running environment. Furthermore, the mechanical robustness of the conductive thread interconnects was assessed across all 12 participant sessions. Despite the strenuous, high-strain 'run to exhaustion' protocol, no signal dropouts, connection failures, or significant data loss attributable to the interconnects were observed. This provides positive preliminary evidence that routing the conductive threads along the garment's seams is a viable strategy for mitigating mechanical strain during dynamic athletic movements.

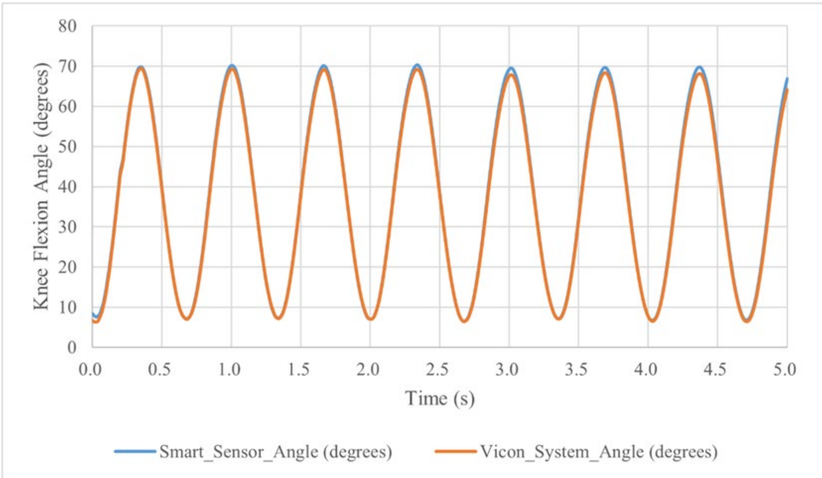

Figure 2. A graph showing the validation of the kinematic sensor. The solid blue line represents the knee flexion angle (in degrees) measured by the smart textile's piezoresistive sensor over a 5-second interval

Analysis of Neuromuscular Fatigue (EMG Data)

The analysis of the EMG data revealed classic signs of neuromuscular fatigue in the F state compared to the NF state. As detailed in Table 1, there was a statistically significant decrease in the Median Frequency (MDF) for both the rectus femoris (RF) and biceps femoris (BF) muscles. The MDF for the RF decreased from an average of 70.1±4.5Hz in the NF state to 55.4±4.7 Hz in the F state ($p < 0.01$). A similar significant decrease was observed for the BF, from 61.5±4.9 Hz to 49.8±5.3 Hz ($p < 0.01$).

Concurrently, the normalized EMG amplitude (%MVC) showed a significant increase with fatigue. The RMS for the RF increased from 42.5 ± 5.0 %MVC in the NF state to 45.9 ± 6.1 %MVC in the F state ($p < 0.05$). Similarly, the BF activation increased significantly from 32.1 ± 4.5 %MVC to 39.5 ± 5.8 %MVC ($p < 0.05$). These results are visualized in Figure 3 a-d. The combination of a downward shift in signal frequency (decreased MDF) and an increase in signal amplitude (increased RMS) provides strong, corroborating evidence of the development of localized muscle fatigue in the primary leg muscles.

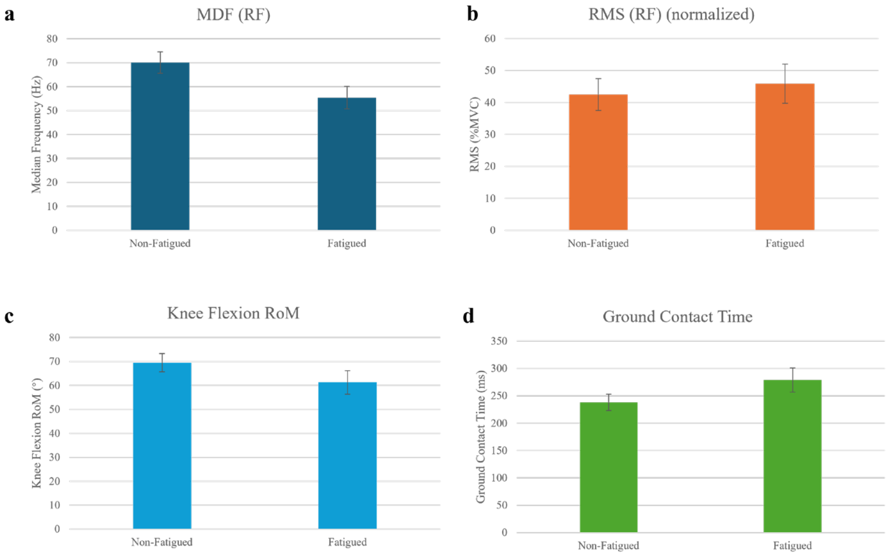

Figure 3. Bar charts comparing the mean values (± standard deviation) of key parameters between the Non-Fatigued and Fatigued states. (a) Median Frequency (MDF) for RF. (b) Normalized RMS amplitude (%MVC) for RF. (c) Knee Flexion Range of Motion (RoM). (d) Ground Contact Time (GCT)

Table 1. Comparison of EMG and Biomechanical Parameters Between Non-Fatigued (NF) and Fatigued (F) States (Mean ± SD, n=12)

Parameter	Non-Fatigued (NF)	Fatigued (F)	p-value
EMG - Rectus Femoris			
Median Frequency (MDF) (Hz)	70.1±4.5	55.4±4.7	< 0.01
RMS (%MVC)	42.5±5.0	45.9±6.1	< 0.05
EMG - Biceps Femoris			
Median Frequency (MDF) (Hz)	61.5±4.9	49.8±5.3	< 0.01
RMS (%MVC)	32.1±4.5	39.5±5.8	< 0.05
Biomechanical			
Knee Flexion RoM (swing phase) (°)	69.5±3.8	61.3±4.9	< 0.05
Ground Contact Time (GCT) (ms)	238±15	279±22	< 0.01

To investigate potential confounding effects of the variable run durations (Time to Exhaustion, TTE), a Pearson correlation analysis was performed between TTE and the magnitude of fatigue-induced changes (ΔMDF and

ΔRoM). No significant correlation was found ($p > 0.05$, $R^2 < 0.1$), indicating that the sensor system captured a consistent fatigue signature independent of the absolute run duration. Furthermore, despite the sample size of n=12 the magnitude of the observed differences was substantial, with large effect sizes (Cohen's d > 1.5) recorded for both EMG median frequency shift and knee range of motion reduction.

Analysis of Biomechanical Alterations

The kinematic and gait data revealed significant alterations in running mechanics as participants transitioned from the NF to the F state. The range of motion of the knee during the swing phase decreased significantly, from an average of 69.5±3.8 degrees to 61.3±4.9 degrees ($p < 0.05$). This indicates a "stiffer" or less dynamic leg swing as fatigue developed.

Furthermore, a highly significant increase in ground contact time (GCT) was observed. The average GCT increased from 238±15 milliseconds in the NF state to 279±22 milliseconds in the F state ($p < 0.01$). This represents an average increase of nearly 17%, suggesting a change in running strategy, likely to maintain stability and compensate for reduced muscular force-generating capacity. These biomechanical changes, summarized in Table 1 and Figure 3, occurred in concert with the observed indicators of neuromuscular fatigue.

DISCUSSION

Critically, this study did not just identify muscular fatigue but also quantified its direct consequences on running mechanics. While the GCT was measured by the laboratory Vicon system and not the textile itself, its strong correlation with the textile-derived fatigue data (EMG and knee RoM) is a significant finding. In a non-fatigued state, proficient runners aim to minimize GCT to improve running economy and reduce braking forces. As neuromuscular function degrades, athletes often unconsciously increase GCT. This can be interpreted as a compensatory strategy to allow more time for force generation and to enhance stability in the face of decreased muscular control and proprioceptive acuity. This finding aligns with studies using laboratory force plates which have linked fatigue to increased GCT and vertical loading rates, factors often associated with an elevated risk of stress-related running injuries.

The concurrent decrease in knee flexion range of motion during the swing phase is another important biomechanical marker of fatigue. This suggests a "stiffening" of the running gait, where the leg is not brought through its full, efficient range of motion. This can be due to several factors, including reduced power from the hip flexors and hamstrings, or a protective mechanism to reduce the load on fatigued knee extensor muscles during the subsequent stance phase. A potential mechanistic link between these two gait changes can be hypothesized from the specific muscle fatigue observed. The fatigue of both the rectus femoris (RF) and biceps femoris (BF) compromises the neuromuscular system's ability to generate rapid propulsive force and absorb landing impact. The observed increase in GCT is likely a direct compensatory strategy for this loss of peak power, allowing the fatigued muscles a longer duration to generate the necessary impulse to maintain pace. Concurrently, the reduced knee flexion RoM during the swing phase suggests a 'stiffer' and less efficient leg recovery. This is likely a direct *effect* of reduced power from the fatigued hip flexors (including the RF) and hamstrings (BF) to dynamically move the limb. Therefore, rather than one gait change causing the other, it is more probable that both the increased GCT and reduced Knee RoM are simultaneous, compensatory *effects* stemming from the same root cause: the declining force-generating capacity of the primary fatigued muscles. By capturing the muscular cause (EMG changes) and a primary mechanical effect (knee kinematic changes) with a single, integrated system, a much richer picture of the athlete's state emerges. The concurrent significant alterations observed in these textile-derived metrics alongside externally measured gait parameters (i.e., GCT) further validates the system's utility for holistically assessing fatigue. Specifically regarding the sensor configuration, we acknowledge that clinical standards typically recommend placing reference electrodes strictly over bony prominences to minimize volume conduction. However, in a wearable compression garment, placing sensors over rigid bony landmarks (e.g., the patella or fibular head) can cause significant discomfort and skin abrasion during dynamic running. Therefore, the distal ITB was selected as a necessary trade-off to prioritize wearer comfort and sensor-skin contact stability. Crucially, the validity of this design choice is supported by our validation data (R=0.96 against standard electrodes), which indicates that this placement provided a high-fidelity signal without significant crosstalk that would compromise fatigue detection.

The implications of this research are twofold. From a textile engineering perspective, it confirms that a combination of screen-printing and piezoresistive polymer integration can create a durable and accurate multimodal sensor system within a standard performance garment. The sensor's stability under fatigued (sweaty) conditions (R=0.97, no significant drift) contradicts the typical hygroscopic instability of textile sensors. This robustness is attributed to a triple-layer protection strategy: First, the silicone encapsulation provided a hermetic seal (Ionic Isolation), preventing conductive sweat ions from shorting the piezoresistive element. Second, placing the sensor on the exterior of the textile leveraged the fabric as a thermal insulator against direct skin heating while exposing it to convective cooling (Thermal Buffering). Third, and critically, the high-compression nature of the base garment meant the fabric was continually pre-tensioned against the skin. This macroscopic Mechanical Locking dominated the mechanics of the sensor-fabric interface, effectively overriding the microscopic hygroscopic expansion (swelling) of the wet fibers that would otherwise alter the sensor's baseline resistance. Thus, the 'zero-offset' was maintained not by the absence of environmental stress, but by the mechanical dominance of the garment's compression. The challenges of washability, long-term signal stability, and mass-manufacturing remain but this work provides a solid foundation for future development. From a sports science perspective, the potential is immense. A coach or athlete could use such a system to get real-time feedback during a training session, identifying the point at which technique begins to break down due to fatigue. This allows for the immediate adjustment of training volume or intensity to prevent overtraining and reduce the risk of injury associated with poor form. Longitudinally, the data could be used to track an athlete's fitness progression, as a fitter athlete should be able to maintain efficient biomechanics for longer durations or at higher intensities. However, it is important to acknowledge the limitations of this study. The sample size (n=12) was relatively small, and a formal a priori power analysis was not conducted. However, the magnitude of the observed fatigue-induced alterations was substantial, with large effect sizes (Cohen’s $d > 1.5$) recorded for both EMG and kinematic metrics. These high effect sizes indicate that, despite the limited cohort, the study possessed adequate statistical power to detect the significant physiological changes characteristic of the 'run to exhaustion' protocol. Furthermore, the sample consisted of only male recreational runners. This exclusion criteria was applied to control for the confounding effects of the menstrual cycle on core body temperature and ligament laxity, which could introduce variability in both the piezoresistive sensor response (due to thermal drift) and kinematic fatigue patterns. By restricting the cohort to males, we aimed to isolate the sensor's performance variables in this initial validation phase. Further research is needed to validate these findings in elite athletes and female populations, who may exhibit different fatigue responses. Furthermore, the 'run to exhaustion' protocol resulted in high inter-subject variability in total exercise duration. While this variability—combined with the reliance on subjective 'voluntary exhaustion' rather than metabolic gold standards (e.g., blood lactate)—implies potential differences in dominant fatigue mechanisms, a post-hoc analysis revealed no significant correlation ($p > 0.05$) between total run duration and the magnitude of the measured degradation. This lack of correlation suggests that the smart textile system successfully captured a consistent 'functional breakdown' signature (i.e., biomechanical failure) that is independent of the specific physiological duration, metabolic pathway, or psychological motivation required to reach that state. The study was also conducted in a controlled laboratory setting on a treadmill. The translation of these results to overground running in variable outdoor conditions, where factors like terrain and wind resistance play a role, requires further investigation. Moreover, a significant limitation of the current kinematic sensor modality itself must be acknowledged. The piezoresistive sensor, as implemented, provides a one-dimensional analysis restricted to sagittal plane knee flexion. This is a severe simplification of a complex 3D movement. Consequently, this system is blind to potentially critical compensatory strategies that manifest in the frontal plane (e.g., dynamic knee valgus) or transverse plane, which are also well-established indicators of neuromuscular fatigue. Future iterations should aim to integrate multi-axis inertial sensors or develop alternative textile sensing structures capable of providing a more comprehensive 3D kinematic analysis. A pertinent limitation is the reliance on static calibration for dynamic capture. The high correlation (R=0.98) observed contradicts the typical behavior of raw piezoresistive materials, which often suffer from significant hysteresis. This performance can be explained by the mechanically constrained environment of the sensor: the tight integration with the compression fabric likely pre-tensioned the sensor, operating it in a more linear region of its resistance-strain curve and minimizing buckling-induced non-linearities. Nevertheless, for higher-frequency movements (e.g., sprinting or impact vibrations), this static calibration would likely fail, necessitating dynamic compensation algorithms (e.g., LSTM or inverse modeling) in future iterations. Regarding the sensor configuration, the decision to place the reference electrode on the distal ITB deviates from anatomical gold standards to accommodate the constraints of a wearable form factor. However, our validation results (Section "Sensor System Validation") showed no significant increase in baseline noise or decrease in signal correlation compared to standard configurations. This suggests that in dynamic, field-based monitoring, prioritizing the mechanical stability of the sensor interface (to reduce motion artifacts) is as critical as anatomical positioning. Thus, the distal ITB proves to be a viable and robust reference site for this specific application. Future iterations of the garment should incorporate a reference site over an electrically neutral landmark, such as the iliac crest, even if it requires more complex wire routing. Finally, a primary limitation of the current study is that the sensors were only validated for accuracy in a single-use session. The long-term durability and performance after multiple wash-and-wear cycles—a core value proposition for functional e-textiles—was not investigated. This is a significant omission, and it must be stressed that the current findings validate only the *acute* accuracy, not the *practical viability or robustness*, of the textile sensor fabrication method. This remains a critical and unaddressed gap that must be the immediate focus of future research. Future work should aim to address these limitations and explore the use of machine learning algorithms to process the multi-modal data streams, potentially creating predictive models that can forecast the onset of critical fatigue before significant biomechanical degradation occurs.

CONCLUSION

This research successfully designed, fabricated, and validated a smart textile system integrating neuromuscular and kinematic sensors for the analysis of running fatigue. The study provided high-quality data demonstrating that the onset of fatigue is characterized by significant neuromuscular changes quantified by the textile (decreased EMG median frequency, increased EMG amplitude) and concurrent biomechanical alterations. Specifically, the smart textile system successfully quantified the reduction in knee range of motion. While the system itself does not monitor foot-ground interaction, this knee kinematic metric was shown to be strongly correlated with the prolonged ground contact time observed by the external Vicon system.

The strong validation against gold-standard laboratory equipment confirms the scientific reliability and accuracy of the textile-based approach. The results provide a compelling case for the use of multi-modal smart garments as powerful analytical tools in sports science. By providing comprehensive, real-world data on the interplay between muscular effort and mechanical execution, this technology has the potential to move beyond simple activity tracking and fundamentally change how athletes train, recover, and prevent injuries, bridging the long-standing gap between the sports science lab and the field of play.

Author Contributions

Xuemin Han and Dong Zhao designed the study; all authors conducted the study; Hongxia Yang and Long Yu collected and analyzed the data. Xuemin Han and Gaihong Yang participated in drafting the manuscript, and all authors contributed to critical revision of the manuscript for important intellectual content. All authors gave final approval of the version to be published. All authors participated fully in the work, took public responsibility for appropriate portions of the content, and agreed to be accountable for all aspects of the work in ensuring that questions related to the accuracy or completeness of any part of the work were appropriately investigated and resolved.

Conflicts of Interest

The authors declare no conflict of interest.

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Availability of Data and Materials

The datasets used and/or analysed during the current study were available from the corresponding author on reasonable request.

Ethics Approval and Consent to Participate

This survey was conducted in compliance with Ethics Committee of Hainan University. Participants were informed of the study's purpose and data usage prior to participation, and responses were collected anonymously. No personally identifiable information was stored.

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Not applicable.

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