REVIEW ARTICLE

Continuous Monitoring and Modeling Contractility of Skeletal Muscles in Motion

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Abstract

Continuous monitoring and analysis of skeletal muscles' contractility have been extensively associated with sensing and bio-signal processing technologies and increasingly demanded by applications in the fields of sports, control and interaction, rehabilitation and medical care. While most existing approaches are confined in isometric studies in clinics or laboratories, researchers have been devoted in recent years towards continuous monitoring and analysis of skeletal muscles' contractility in motion. This paper aims to provide an overview of current status of non-invasive sensing technologies for monitoring skeletal muscles' activation, up-to-date findings on observing and characterizing the force-length and force-velocity relationships, and various existing activation-contractility models. In addition, this paper evaluates various sensing technologies for muscle activation, indicates challenges for bio-mechanical modeling on activation-contractility, and makes recommendations on future developments in continuous monitoring and analysis of skeletal muscles' contractility in-motion.

1 Introduction

As the motors of the human motions and stabilizers of joint positions, skeletal muscles have attracted great interests of researchers on their contractile properties and characteristics. The bio-mechanical analysis of the contractility has been constantly required and extensively incorporated in the areas of rehabilitation and medical care, control and interactions, as well as sports training [1-5]. However, direct measurements of muscle contractility are impossible, and inverse dynamics analysis only provides a net output of moment by all surrounding skeletal muscles around a joint. In order to determine the tension of individual muscles, using a variety of sensing technologies to link the muscle activation to contractility based on bio-mechanical models (Figure 1) remains the only feasible choice.

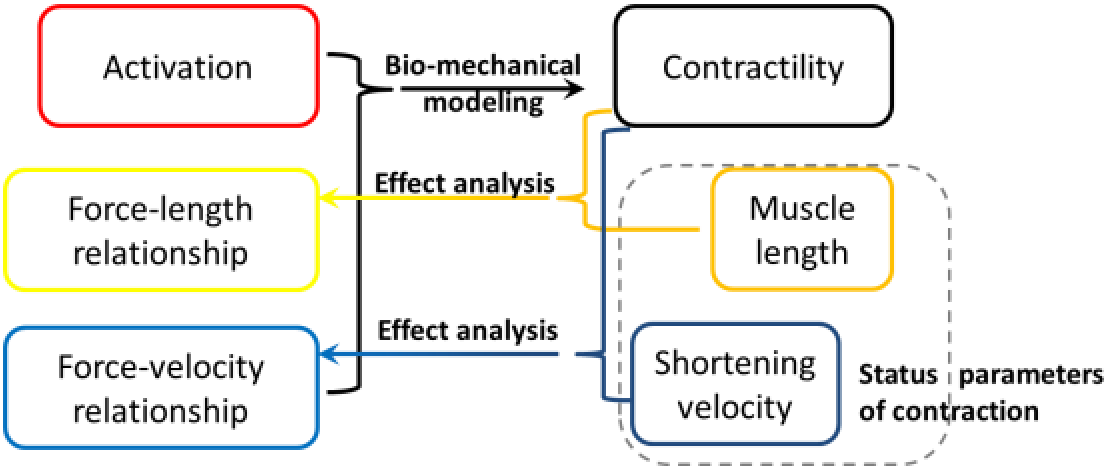

Figure 1. Typical Hill-type modeling process for activation-contractility relationship

Hence, this review delivers a conflation of up-to-date researches on monitoring the activation of skeletal muscles as well as the determination of force-length and force-velocity relationships, both are essential for the bio-mechanical modeling of activation-to-contractility, and presents the merits and challenges for in-motion monitoring of muscle contractility. Since studies of contraction in impaired muscles, such as neuromuscular diseases and fatiguing contraction, were reviewed somewhere else [6-13]. In this paper we mainly focus on non-fatigue muscle contractions of healthy subjects.

2 Monitoring Muscle Activation

2.1 Classical approaches

Skeletal muscles' contraction is generally accompanied by three typical kinds of bio-physical behavior, namely the polarization-depolarization of sarcolemma, vibration of tensioned muscle fiber, shortening in length while expansion in cross-sectional area of the muscle fiber. Accordingly three types of non-invasive approaches for monitoring muscle contraction include electrical (surface electromyography), mechanical (mechanomyogram), and morphological (computed tomography, magnetic resonance imaging, ultrasound, etc.) [1, 14-16]. Over the past decades, efforts have been made to track muscle contraction continuously by correlating bioelectricity of superficial skeletal muscles [3, 17-21], low-frequency oscillation of firing muscle fibers [21-23], as well as changing in specified morphology parameters of muscles during contraction [24-28],with muscular contraction status (mostly indexed by contractile force). One thing should be noted is that, among all these attempts, the majority of quantitative researches are merely limited to isometric contractions. According to Hill's theory of muscle contraction, contractility is affected by activation of skeletal muscle and the status of contraction, apart from the maximum voluntary contraction of skeletal muscle(s). Isometric contractions are convenient because with fixed muscle length or joint position, the so-called 'monitoring of muscle contraction' is in essence the monitoring of indexes of muscle activation, which have been realized by the aforementioned three types of sensing technologies. However, as a successful index of activation for motion, the following aspects of each technology need to be considered: (1) the nature of the activation–force relationship, (2) stability of acquisition for all contraction modes. Further for activation-contractility modeling, elaborations of the influence factors on dynamic muscle contractions shall also be observed.

2.1.1 Surface electromyography (sEMG)

sEMG provides information on individual muscle's bio-physical activity, which has been correlated with contractile force. sEMG collects the action potential through electrodes attached to the skin surface at the middle of the muscle belly [29]. As a composition of action potentials of multiple motor units, sEMG signal has been widely applied for both clinical and research purposes [30]. In particular, sEMG has been extensively used to study the muscular functions and muscles' coordination. The research scope of this area mainly consists of the following four aspects, where the 1st, 3rd and 4th aspects are related to the contractility of skeletal muscles:

1. isometric contraction, maximal voluntary contraction (MVC) and the relationship between sEMG and contractile force [10, 21, 31-34];
2. healthy skeletal muscles' anatomical function during assigned movements or sports, jointly with other synchronized bio-signals [11, 35-38];
3. muscle fatigue [39];
4. computational and clinical studies of sEMG on assisting occupational rehabilitation, including sEMG-to-motion modeling [3, 35, 40-43].

In an oscillation wave-shape, the sEMG signals can be analyzed and interpreted in frequency domain and time domain, the absolute amplitudes ('envelop' or the outline) of the later is related to the intensity of contraction. For a simple comparison analysis, the magnitudes of sEMG representing the gross innervation, with which the skeletal muscle works, avoiding the influences of non-reproducible components of sEMG at different time-intervals [44, 45]. For continuous monitoring purpose, it is preferable to normalize the in-motion amplitude with Maximum Voluntary Contraction (MVC) to index the level of muscle activation [3, 35, 40]. High correlation between the normalized sEMG and force generation has been reported constantly in both linear and nonlinear patterns. A typical relation is monotonic and curvilinear, a higher sEMG is needed for a further unit increase in contractile force [21, 46, 47]. For static tests, when both sEMG and force are normalized to their respective maximum values, some muscles tend to show an apparent linear EMG-force relationship [48, 49]. The investigation of such relationships is important since sEMG has been used for assisting EMG-to-motion applications based on biomechanical models [40, 42, 50]. However, there are two pertinent problems, especially for in-motion application. The first is that sEMG signals are sensitive to the size, number and firing rate of muscle units, which undermines the role of sEMG as representative index for the intensity of selected muscles' contraction, not to mention that sEMG signals are strongly influenced by conditions of detection, such as the skin humidity, contact resistances, both are difficult to maintain constant in fierce motions. The dependence on the instrumentation and acquisition procedures may have influences, which have been extensively discussed elsewhere [51-53]. Secondly, a recent work shows that the sEMG normalization is limited and inaccurate especially for high-velocity dynamic tasks, which implies that the optimal normalization methods should be muscle and task-dependent [46]. Because concentric contractions shorten the muscle length, the location between muscle and sEMG electrodes may be changed. Moreover, the status of muscle contraction, such as length and shortening velocity, also affects the sEMG signal [54, 55]. Up to now, the quantitative studies have been limited on isometric contractions for limited time duration. Hence, although sEMG is now the most common noninvasive approach to monitor skeletal muscles' activation, it still confronts several challenges as an index of muscle activation in motion.

2.1.2 Mechanomyography (MMG)

MMG signals represent the vibrations of active muscle fibers during contraction, which can be detected by piezoelectric sensors [9, 56], microphones [57, 58], accelerometers[47, 59, 60] and laser distance sensors [61, 62] .The oscillations, reported as prominently influenced by the global firing rate of motor units [63-65], reflect the mechanical counterpart of the electrical activity of the motor unit as measured by sEMG [19]. Compared with sEMG, MMG signals cover a wider physiological range of motor units, even the underlying muscles, only that the waves oscillate as discrete bursts rather than continuous tones [66]. Furthermore, the placement of MMG sensors is not required to be precise or specific [67], and MMG signals are not influenced by changes in the skin impedance and sweating [68].

MMG studies on muscle activities are numerous, including characterization of neuromuscular disorders [6, 69, 70], development of prosthesis and/or switch control [64, 65, 71, 72], activity of motor units[73-76], examination of mechanical properties during exercises [77, 78], and rehabilitation systems [79]. Temporal and spectral components of MMG signals with difference levels of contraction have been employed to determine muscle strength and stiffness. Studies of MMG versus isometric torque of human elbow indicate that the relationship between MMG amplitude and isometric torque is linear for lower-strength subjects and cubic for higher-strength subjects [80-82]. It was reported that the RMS of MMG decreases at high levels of force due to mechanical fusion of MU activity while the mean power frequency of MMG increases [83, 84].

As an alternative promising monitoring approach for muscle contraction, however, MMG has not been developed fully as it cannot determine the activation level of muscle to reflect contractility. Existing studies on MMG are confined to a small sample size of healthy population [64, 85]. Recently, A recent study combining sEMG and MMG has facilitated new understanding of the electromechanical coupling of skeletal muscles [21, 22, 75, 86]. Noise contamination is also a major barrier for MMG. Low-frequency (5-100 Hz) MMG is not perfect for in-motion monitoring of muscle contractions, due to the fact that the low-frequency components of MMG are easily mixed with human movements. On the other hand, the high-frequency components of MMG are contaminated by nearby vibrating muscle fibers or environmental noises [68].

2.1.3 Tomographic imaging methods

Skeletal muscle's architecture alters with contraction [87], as muscle fibers shorten and simultaneously change the morphology of the entire skeletal muscle. Imaging methods such as the ultrasound (ultrasonography), computerized tomography (CT), and magnetic resonance imaging (MRI) have all been implemented to detect morphological deformation of skeletal muscles in real-time. Through on-line or post imaging processing, architectural parameters of skeletal muscles can be quantitatively identified, such as the changes in cross-section area [24-26, 88] and muscle volume [27, 28], muscle thickness and fiber pennation [20, 21, 89], through image processing of cross-sectional area of skeletal muscles. Those architectural parameters have been reported as index of contractile force during isometric contraction, and correlations between the morphological parameters and generated joint torque were achieved [15, 20, 21, 90]. Being low-cost, non-ionizing, stable, and available for deep muscles, ultrasonography has been widely applied in researches covering:

1. Correlations among change of CSAs (cross-sectional areas), expansion of muscle size and contractile force have been reported frequently [28, 29, 88, 91-98];
2. Shortening fascicle length and increasing pennation angle were also reported to be highly correlated to skeletal muscle's contraction [99-103];
3. Muscular movement or displacement was detected with in-vivo ultrasound [104, 105];
4. Potential of muscle fatigue evaluation and prosthetic control using the ultrasound has been discussed [106], providing more comprehensive information than using sEMG only.
5. Muscle thickness extracted from CSA [107] have been jointly studied and found correlated with other indexes of muscle contraction [21, 91, 101, 108-110]. Corresponding image tracking algorithms have been developed for in-motion detecting. For relaxed skeletal muscle, muscle thickness was also observed in close relationship with joint angle due to the shortening of muscle length.

The 1st, 2nd, and 5th items are related to monitoring of muscle contraction. In particular, for the biceps brachii (major components of elbow flexors), Hodges [111] found that the muscle thickness has a negative exponential relationship with sEMG signal, i.e., the muscle thickness increases with sEMG almost linearly in low-contraction (<30% MVC) while much slower in high-contraction condition. This observation was confirmed by Akagi [112], Abe [113], and Zheng's group [20, 21], who developed a system to record and analyze ultrasound images, force, joint angle and sEMG simultaneously. However, due to the difficult fixation of ultrasound sensor in dynamic conditions, almost all the studies were restrained in static (isometric) and quasi-static conditions.

Attempts have been made to establish the relationship between the measured muscle size (eg, thickness and/or cross-sectional area) and the level and timing of muscle activation [114]. The level of muscle activation was determined by comparing the size of a contracted muscle to its size during rest. Using measured muscle size from static ultrasound images as an indication of muscle activation, however, has limitations. The level of muscle activation depends not just on a muscle's size, but on initial muscle (fascicle) length, amount of tendon stretch, type of contraction (isometric, concentric, or eccentric), muscle fiber pennation angle, and forces from surrounding tissues [115] [116, 117]. Up to date, there have not been sufficient studies on examining the reliability and validity of ultrasound for quantifying muscle activation during research and clinical practice [118].

2.2 New technologies for monitoring muscle contraction

New sensors and sensing technologies have illustrated great promises for monitoring and analyzing skeletal muscles' contraction. From recent papers published between 2014 and 2018, as shown in Table 1 and Figure 2, tensiomyography, optic sensors, novel ultrasound sensors, piezoelectric sensors and large-deformation strain sensors have been reported.

Table 1 Summary of new sensing technologies for monitoring muscle contractions

References	Type of sensor used	Principles	Remarks
[119-123]	Tensiomyography	Radial muscle belly displacement under electrical stimulus is analyzed for neuro-muscular function of muscle	For evaluating contraction only; Superficial muscle only; Isometric mode only;
[124-128]	LED & photo detector	Light absorption and reflection by muscle fibers vary during contraction; Tissue oxygenation decreases in contraction	Qualitative monitoring of muscle contraction

[129, 130]	Wearable ultrasound piezoelectric PVDF sensor	Muscle thickness increases during contraction	Qualitative monitoring of muscle contraction; Isometric mode only;
[13, 131]	Piezoelectric sensor	Modulus of muscle increases in tension	For analysis and evaluation of contraction only; Lack of dynamic study;
[132-135]	Large-deformation strain sensor	Expansion of CSA of muscle fibers during contraction	Model for activation-contraction; Involving both isometric and kinetic modes.

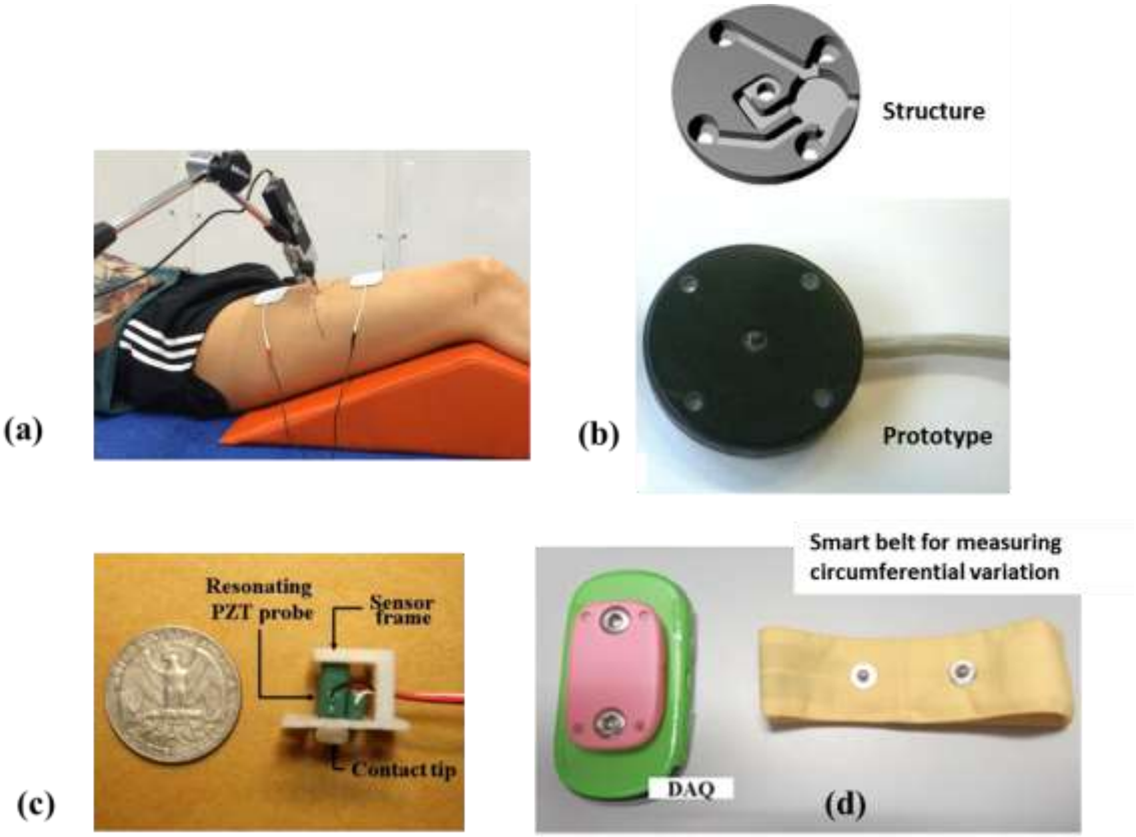

Figure 2. (a) Tensiomyography sensor [121]; (b) electro-optical muscle sensor [125]; (c) piezoelectric sensor [131] and (d) upgraded soft strain sensor [134]

Tensiomyography

Tensiomyography (TMG, sometimes the MC sensors) is a portable non-invasive method to assess in vivo passive muscle contractile properties [136]. Inspired by MMG, tensiomyography uses a high-precision digital transducer placed on muscle surface to capture waveforms integrating parameters such as maximum radial displacement of the muscle belly and contraction time [137, 138]. Compared to MMG techniques, TMG signals are not affected by slight muscle pretension, and thus have a higher signal-to-noise ratio.

TMG has been used for evaluation of muscular fatigue[12], impairment [139], as well as muscular changes/adaptations [140]. Variations of TMG-derived parameters show significant correlation with changes in MVC [123]. TMG has been appreciated by strength and conditioning coaches, physiotherapists, and sport scientists, who preferentially seek accurate and practical assessment methods which do not disturb their professional routines [141, 142]. However, TMG has a number of shortcomings. First, studies of TMG have been confined in isometric mode, no dynamic application is yet possible. Secondly, TMG is not able to assess deep muscles. Finally, congestion due to training may affect the accuracy of TMG data. In summary, due to the lack of in-motion monitoring ability and stability, TMG can serve as an evaluation tool of contractile properties such as muscle fatigue and capability, other than monitoring muscle contraction.

Optical sensors

Optical sensing devices for muscle contraction consist of light emitting diode (LED) and photo detector (photodiodes) [124-126]. The working principle is to measure the change in intensity of back scattered light from skeletal muscle tissue, which is caused by myosin proteins' crystalline properties during contraction. Recent work [124] shows that photodiodes can also derive the tissue oxyhemoglobin absorption from the measured light intensity, i.e., the steady decrease in the tissue oxygenation during ischemia. However, studies on optical sensors are still in initial stage, especially as a tool for assessing muscle contractility.

New wearable piezoelectric ultrasound sensors

Inspired by conventional ultrasound methods, simple and wearable sensing devices have been designed for monitoring muscle contraction by quantifying muscle thickness and active muscle stiffness, respectively. Examples include disk-shaped ultrasound device [109] and piezoelectric sensor-based reasoning device [103]. The latter is more of an evaluation tool of contractile properties while the wearable ultrasound sensor needs to prove its stability of signal in dynamic conditions.

Anthropometric measurement devices

Involving measurements of various dimensional descriptors of human body, anthropometry has been widely used in industrial and clothing design, ergonomics and human fitness evaluation [143]. Muscle size from anthropometric measurement (e.g. limb circumferences) has been found to correspond to the muscularity [108, 144-149], i.e., higher muscular strength is associated with greater limb circumference and vice versa. Since the thickness, identified from the muscle cross-sectional area, has been frequently reported as architectural index for skeletal muscle's contraction [21, 91, 101, 108-110], the change in limb circumference induced by expansion of cross sectional area may be another index.

Correlations between the limb circumference and torque at MVC have been reported, moreover, most of which were obtained at static or quasi-static conditions [132, 145, 150, 151].

The anthropometric measurement renders a number of wearable sensing devices that facilitate long-term continuous monitoring in dynamic conditions. A mechanical armband with steel wires was demonstrated [152] for measuring the circumference of human forearm in-motion, and found an apparent linear relationship between forearm circumferences and grip force, which is in agreement with the strength-size research findings [153, 154]. By using a muscle circumference sensor (with metal wires), Kim[132] proposed a preliminary Hill-based model for human upper arm, elbow torque was predicted from the measured mid-upper-arm circumference with significant estimation error. There was no in-depth studies published supporting the arbitrary replacement of sEMG with measured mid-upper-arm circumference as a new ‘activation level’. Other factors should be further studied, such as the joint position and speed of flexion. Nevertheless, these works have inspired developments of wearable monitoring systems for deformation of muscle during contraction. However, the measurement devices were rigid, interfering with muscle activity. A new type of strain sensors has been commercially available for large repeated deformation up to 60%, high sensitivity and good accuracy [14, 155-160]. They were made from elastic fabrics coated with elastomer/carbon nano-particles composite. Wang [133] used a measurement device with these fabric strain sensors and studied the relationship between upper arm circumferential strain and elbow flexion, in isometric, isokinetic and isotonic flexions. He has obtained empirical relationships between the circumferential strain and contraction torque [133], in addition, a biomechanical model for kinetic flexions was proposed based on the observations of force-length and force-velocity relationship [134]. The model was validated for isokinetic contractions for moderate speeds. As the derivation of force-velocity relationship was based on slow or median speed, lack of experimental evidence on contraction at high-shortening speeds (>10 m/s or 450°/s), there is still a question unanswered, that is, how can the circumferential strain be linked to the muscle activation, especially in high-speed dynamic conditions.

It is noted that these anthropometric studies have focused on the activation based on measured contraction-induced circumference strains and the activation-contraction model even in very early stage [132, 134].

In general, studies of muscle activation using conventional technologies (EMG, MMG and tomographic imaging) are still restricted in static/isometric contractions instead of in-motion, due to aforementioned drawbacks impairing the signal stability in dynamic conditions. The new technologies have not been tested yet for long thus no sufficient cases have been reported. TMG has not been used for in-vivo monitoring of contraction. The optical sensors are immune to electric and electronic disturbance but affiliated bulky and heavy modem and wires prevent them from in-field applications. The piezoelectric ultrasound sensors need to prove their signal stability in dynamic conditions. The anthropometric measurement devices is convenient for sports,extracting the strains of individual muscles could be a challenge, however. Meanwhile, how the circumferential strain indexes muscle activation in kinetic contractions should be further studied.

3 Biomechanical Models of Activation-Contractility

Although over 80 years have passed since Hill [161] first revealed his biomechanical insights into muscle contraction, quantitative modeling of contractility has been mainly limited by phenomenological implementation of various Hill-based models. Huxley [162] further introduced the dynamics of cross-bridge cycling into the contractile element (instead of a black box in Hill's) and successfully reproduced fast-twitch muscles, which could not be explained by a classical Hill model [163, 164]. However, Huxley's consideration was claimed too computation-time-consuming for use in musculoskeletal modeling [165, 166], due to the complex mathematical formulation. Meanwhile, recent works show that muscle tensions predicted by both Hill and Huxley models are within the same range [165, 167]. Hence, though Huxley's model provides more realistic patterns of muscle contraction, it is economical and reasonable to use a simpler numeric implementation based on Hill type models.

Apart from the activation talked above, the influences of status parameters on the contractility shall be determined, i.e., the force-length and force-velocity relationships. Moreover, the acquisition of the status parameters is addressed, by reviewing sensing technologies on joint angle measurement. All above are essential to complete a Hill-type biomechanical modeling of activation-contractility in motion.

3.1 Hill-type biomechanical models

Once the muscle activation is determined experimentally, the next step shall be to build an activation-to-contraction conversion linkage, commonly known as the biomechanical modeling. As first released in Hill's macroscopic studies of skeletal muscles' contraction [168, 169], the muscle tension has a hyperbolic relationship with shortening velocity [161, 170, 171]. This finding was then extended and expressed by Zajac [172] and Winters [173, 174] in a four-element one-dimensional Hill-type model:

$$F_{CE} = \alpha(t) \cdot f(v) \cdot \left[f(l) \cdot F_{Max} \right]$$

The contraction force of muscle fiber, F_{CE}, is a function of MVC at current muscle length,scale factor for the activation level, $\alpha(t)$, and normalized factor of shortening velocity, $f(v)$ [172, 175-177]. The functions $\alpha(t)$, $f(v)$ and $f(l)$ can be in diverse forms with particular parameters, to be specific to different skeletal muscles [174, 175, 178, 179].

These models have been used extensively for assessing skeletal muscle characteristics [43, 49, 178, 180], contractions and movements [41, 42, 175, 181, 182], analyzing neuromuscular-related diseases and rehabilitation [183-190]. In particular, with sEMG derived activation levels, Hill-type models have been proven effective for emulating muscular behavior [35, 191]. However, the accuracy and reliability of in-vivo muscle forces predicted by these models remains unknown, due to the lack of suitable implanted transducers.

Microscopic muscle fiber models are established with bio-physical tuning parameters that predict muscle characteristics and contractions quite well [192-196]. Though relying on intrinsic properties likewise, namely the construction of activation from bio-signals, the force-length and force-velocity relationships, the macroscopic muscle models, however, are commonly based on experiments and phenomenology [197-200]. The adoption of microscopic sarcomere model to whole muscle or straight application of existing macroscopic model for one kind of skeletal muscle to another, have been reported to induce huge errors in prediction of real contractility, especially for movements at the low or high ends of speed [201]. The reason lies in the fact that the real structure of skeletal muscles and recruitment patters of slower and faster motor units in muscle make the contractile properties largely indescribable [173, 201-205]. This naturally requires task-dependent observations on the determination of parameters in the models [206-208]. Moreover, since muscles are normally surrounded by other tissues, the impedance caused by connective tissues and bones [209, 210] bring some difficulties in modeling.

In summary, to construct a Hill-type biomechanical model for in-motion monitoring of contractility, it is better to determine the core elements of a Hill type muscle model through isolated designed experiments with selected conditions. Hence, the next subsections will cover the research status on determining the two core elements, that is, the force-length and force-velocity relationships.

3.2 Force-length relationship

In the human musculoskeletal system, the tension and status of contraction (i.e., the length and shortening speed) of skeletal muscles are alternatively represented by corresponding joint torque, joint position (or joint angle) and joint angular velocity, respectively [211, 212].

Experimental studies have been conducted on sarcomere, fibers and whole muscles from various animals, mostly cat, frog, rabbit and rat, and in isometric contraction mode [213, 214]. In the isometric mode, the contraction force of muscle was length dependent while velocity was zero and maximum contraction incurred, the same did maximum activation. For muscular-skeletal systems, the force-length relationship is replaced by torque-angle relationship, which differs from the biological force-length relationship since it incorporates the effect of moment arm of skeletal muscles to the joint. There is definitely a difference in optimum muscle length for concentric and eccentric contractions, respectively, as reported by Melo et.al [215] in knee flexion and extension studies. Moreover, the force-length relationship was found to be activation-dependent [216], which is in consistence with Hill's theory.

The first qualitative descriptions of force-length relationship were associated with theoretical considerations on the interaction of actin and myosin, known as sliding filaments theory [192, 217] Since then, a variety of force-length relationship has been proposed. The force-length relationship for the whole muscle is the easiest to obtained from experiments and thus discussed frequently [115, 173, 209]. However, most derived relationships were empirical, based on best data fitting without underlining biomechanical or physiological analysis. A theoretical force-length relationship is more difficult because it should incorporate a complex combination of properties of sarcomere, tendon, and muscle unit, as well as architectural particularities and history of contraction[213, 218-220], not to mention the controversy over variation of muscle length after strength training. Until recently, only one purely physiological model was presented [221], where the force-length relationship was parameterized by using the geometry of internal muscle structures.

3.3 Force-velocity relationship

The force-velocity relationship was generally obtained from various activities at maximum activation, such as cycling [222-224], vertical jumps [225-229], treadmill training [230, 231], leg press [232-234], arm and upper body movements [223, 224, 227, 235]. The standardization and observation of force-velocity relationship are essential not just for routine tests but also for biomechanical modeling. One thing should be pointed out is that the 'force' used by these researchers is either an index of load/resistance or contractility deducted inversely from devices, while the 'velocity' is not exactly the shortening velocity of muscle but a speed of the motion, somehow linked to the skeletal muscles' shortening velocity. For example, recent researches use mean force exerted onto the ground and the mean velocity of the mass center to establish the force-velocity profile in the vertical squat jump [236, 237]. In the human muscular-skeletal system, the joint moments and angular velocity appears to possess a hyperbolic relationship, similar to the original force-velocity relationship for single muscle fibers [238], referred as the joint torque-angular velocity properties. For individual movements, the force-velocity relationship can be determined in isokinetic training mode, by dynamometers, such as Biodex™.

While the force-velocity relationship of isolated muscles has been known to be hyperbolic [161], multi-joint functional tasks typically reveal strong and approximately linear patterns [225, 239, 240]. A linear force-velocity relationship has been observed in the squat [241], leg press [240], free and loaded vertical jumps [225, 236], cycling [222, 230], treadmill running [230, 231], arm cranking [223], in both bench presses and bench press throws[235, 242], and during rowing [243]. Despite the experimental facts, the mechanisms of non-hyperbolic relationship in multi-joint movements are still not clear. As a consequence, no physiological model has been proposed. In the classical Hill-type biomechanical models, the force-velocity and force-length relationships are independent to each other, however, the interaction item between them has been reported [244], which needs investigations in future study.

According to Hill's theory, the best condition to identify and determine the force-velocity relationship is to maintain the maximum activation level, for kinetic modes, 'maximum' means to move as rapidly as possible regardless of the restraints/resistance, which, however, cannot be fully satisfied by maximum movements/tasks in practice. Behm [245] claimed after studying ankle dorsiflexion isokinetically that the "intent to move quickly" is the only important factor for producing accurate velocity-strength relationship. Although numerous isometric studies have constantly shown correlations between the contraction force and activation indicators, activation itself actually represents the level of output power in Hill's model. Recent studies have shown difference in the force-velocity relationships obtained by using different external loads [246, 247]. Furthermore, different shortening velocity also varies the observed force-length relationships [248, 249]. There is always a challenge that although it's long believed that activation, force-length and force-velocity properties are mutually independent, a crossed instead of separated consideration of activation, length and shortening velocity shall be done during Hill type modeling, where combined effect of the three mentioned factors on scaling the output contractility require s further research investigations.

3.4 In-motion angle measurement technologies

As elaborated above, the observation of the key status indicators of muscle contraction, i.e., the length of muscle fiber and the shortening velocity, rely on the detection of joint angles. A brief review of in-motion determination of joint angles base on various technologies is to be presented in this section. More detailed reviews can be found from reference [250-252].

Inertial measurement units (IMU)

For posture measurement, most conventional and common solution is using the inertial measurement units (IMU), which has been implemented either as a stand-alone sensing device or integrated in smart phones. The IMUs can measure angular velocity, acceleration and the magnetic field vector in their own 3D local coordinate system (Figure 3.a). Strap-down integration [250, 256] of angular velocity is used as a preliminary estimate of the displacement, the drifts of which are corrected based on a number of Kalman-filtering algorithms [257, 258]. The combination of multiple IMUs placed on body segments around a joint provides the joint angles [253, 259-261]. However, the measured acceleration and magnetic field vector are disturbed by impact on ligaments and presence of magnetic objects, lowering the accuracy of displacement or orientation. Furthermore, for accurate measurement of joint angles, a multi-IMU system is needed, which often undermines the freedom of movement in motions.

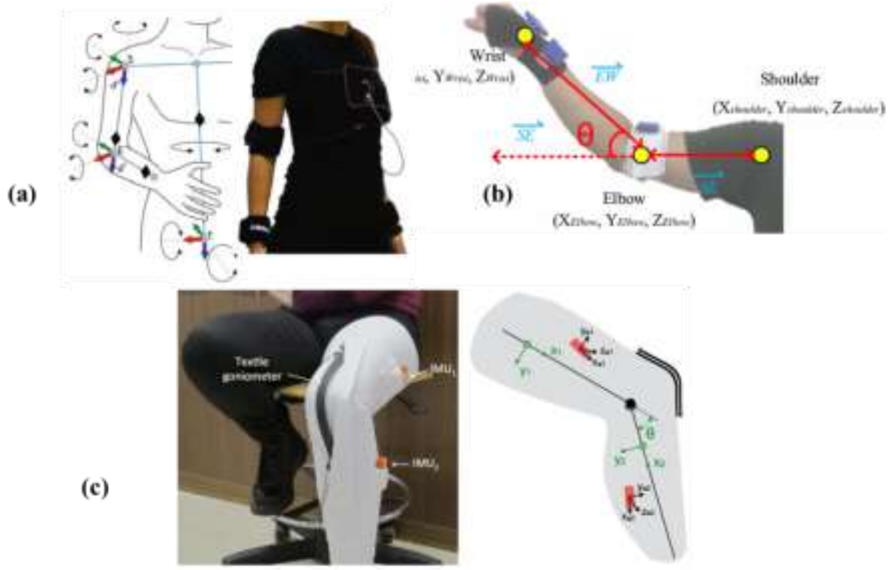

Figure 3. Reported technologies on joint angle measurement/posture recognition: a) IMUs[253] b) optoelectronics sensors[254] c) fabric-based (KPF) [255]

Multi-camera motion capture system

The motion-capture system based on optoelectronic sensors has been used for visual assessment in interaction, physical therapy or rehabilitation [251, 252]. With or without makers on major limbs, optoelectronics sensors work with cameras and 3D post-processing vision system (either contrast based or depth based) to track joint angles through limb orientation and motion of body segments (Figure 3.b), only for major limbs, however. Marker-based vision-capture systems are accurate and reliable, conventionally referred as benchmark [254, 262, 263]. However, they are costly, need professional calibration and strict-conditioned circumstance. Environmental noises in captured images due to occlusion, self-occlusion, and unconventional body postures can induce wrong limb-identifications [264].

Soft goniometers

Recently, soft goniometers such as textile-based wearable sensors, have been working with or without the aforementioned other three types of sensors [265, 266]. Conductive elastomer coated fabrics [267, 268] and knitted piezo-resistive fabric [255, 269-271] have been studied for movements/ postures recognition [268, 269, 271] and joint angle measurement [272], due to the merits of high compatibility with in-field activities (Figure 3.c). Correlations between knee angle and resistance change were observed and characteristics of gait cycle can be accurately identified, with a mean error of <3% [273, 274], comparable to that of commercial IMUs. For simpler on-and-off applications, thresholds have been set to evaluate the range of motion, for instance, whether the target range was achieved [275]. For upper-limbs applications, gloves, sleeves and shirt have been developed based on those soft sensors. With an angle measurement error equal or less than 8%, those garments give reliable identification of static posture of hand, arm and shoulder [276]. However, these prototypes perform poorly in transient measurements, due to the drift in angle-resistance curves affected by stretching speed of the sensing area, as well as in the recovery time prior to the second use [277]. Up-to-date, effort has been reported in optimization of device design and arrangement of sensors, and in employment of IMUs in order to improve the accuracy of measurements [272]. Although some angle-sensing gloves and shirts have been demonstrated for providing feedback for people with central nervous system lesion in therapeutic exercises [7, 267, 269, 278, 279], further research is required to enhance their accuracy, reliability as the angle measurement, diagnosis and rehabilitation tools.

4 Applications

Up-to-date, sEMG has been the only widely used instrumental tool to construct activation of skeletal muscle contraction due to its biophysical nature. Generally, the sEMG-based activation-contraction models have continually been incorporated in studies of prosthetic/supporting robotics, many times merely based on EMG interpretation and pattern learning [280-283], thus will not be elaborated here. By studying how much muscle force is being produced or to be produced for rehabilitation and medical intervention evaluation purposes, therapists can set safe limits in their therapies, meanwhile patients can learn to adjust force production to fulfill designated actions. However, these studies are still confined in static condition/isometric mode of contractions (Figure 4.a). Manal [284] successfully predicted the ankle moments in isometric plantar flexion and dorsiflexion with a tuned sEMG-driven Hill model. A system for quasi-dynamic monitoring of ankle moments and achilles tendon force was also preliminarily presented, consisting of electro-goniometers and EMG sensors (Figure 4.c). Shao [50] from the same group applied the EMG-driven Hill model for four stroke patients and predicted ankle moment during stance with an acceptable RMS error of between 9.7%~14.7%, conforming the model's consistency and effectiveness as rehabilitation therapies to assess intervention. Apart from isometric contractions, Koo [285] and Pau [286] tested their sEMG-driven Hill models in the other way, by comparing elbow joint trajectory predicted with externally measured during isotonic elbow flexions (Figure 4.b). It was claimed that the discrepancy between which was due to muscle activation constructed from sEMG signals in dynamic conditions. More work related to the above two perspectives was reviewed by Biewener [191], suggesting better sensing technologies to further improve the accuracy of Hill-based activation-contractility models among multiple tasks.

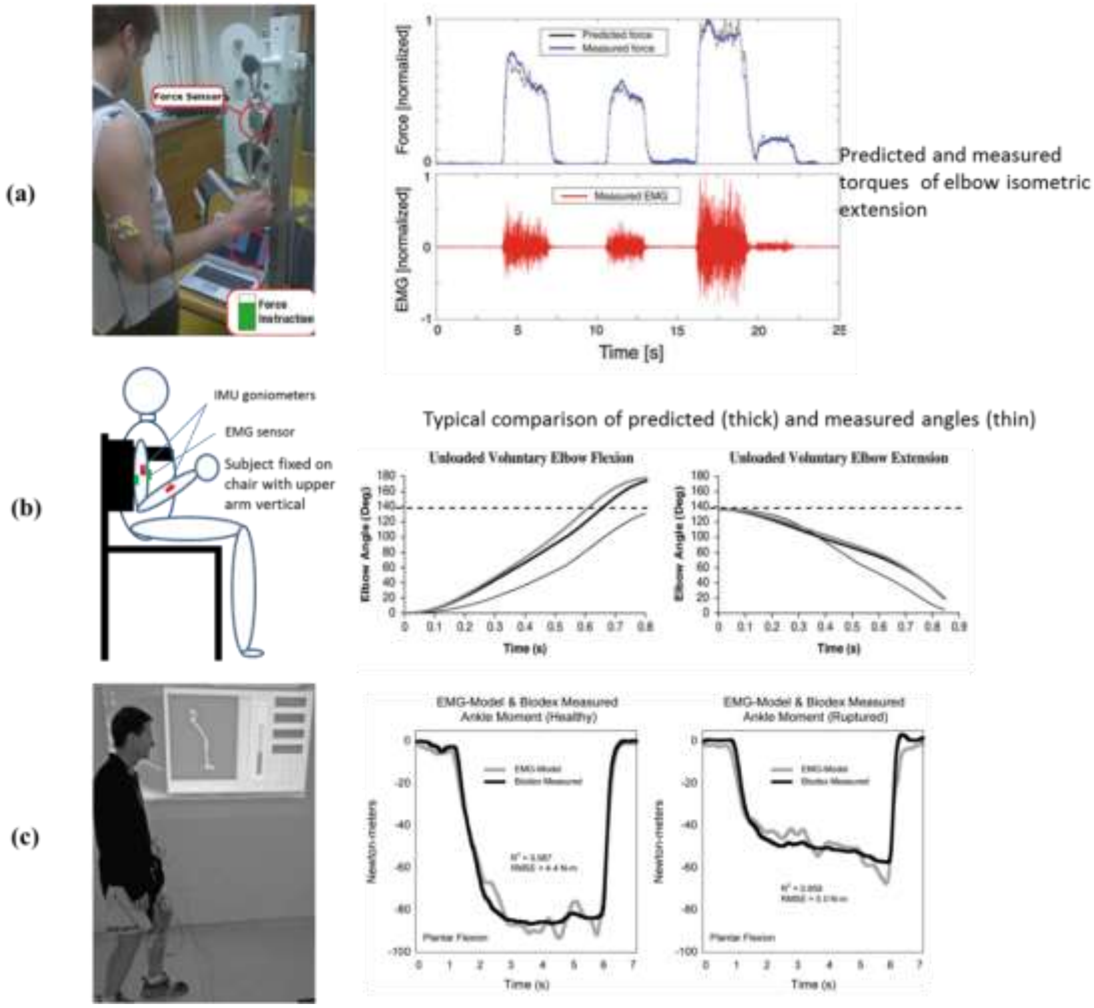

Figure 4. (a) Prediction of toque based on sEMG-force Hill model during isometric contraction[287]; (b) Elbow joint position predicted based on Hill type EMG-force model in isotonic contractions [285]; (c) A proposed system with sEMG and position measurements for in-motion estimation of ankle moment [284]

Due to the limitation of current sensing technologies of activation in dynamic circumstances, no published research has been able to track the muscular tensions in sports and field training, even in very simple specific or chosen tasks. Ligament orientation and joint positions were captured by cameras, Louis [288] studied the a common clinical routine of reach-grasp movements, only able to compare muscle tension determined between different EMG-force models. In a most frequently cited work, Lloyd [41] illustrated general Hill-type modeling, used a modified EMG-driven model to calculate knee moments in crossover cuts and straight runs, and validated the model by comparing the calculated torque with that obtained from inverse dynamics. In this work, joint position was accessed by electro-goniometer. Langenderfer [42] used tuned EMG-driven Hill model in isometric flexions for determining contractility of individual skeletal muscles in fore and upper arms, but only able to compare net moment around elbow joint between measured and predicted. A most recent dynamic application of the EMG-driven Hill model was revealed by Lee [289] on predicting tension of goat gastrocnemius muscles during walking and running, with the RMS error observed as high as 32%. In this work, muscle length and its time rate of change were obtained by lab-made ultrasound sensors.

The reviewed applications above adequately and representatively reflect the current research gaps in monitoring muscle contractility based on activation-contractility modeling: First, although sEMG has been proven frequently effective to assess muscle activation during isometric contractions, it is far way from being reliable in dynamic conditions. As such, additional or alternative descriptions of activation based on other sensing technologies are required; Secondly, simultaneous measurements of activation and contraction status (muscle length and shortening velocity) have been proposed in order to achieve a complete activation-contractility modeling for dynamic contractions, which, however, have not been truly realized until now. Therefore and thirdly, inconsistency of predictions from a model was reported for a diverse range of tasks now and then. Finally, there is still lack of direct or indirect sensing approaches for verification of the muscle tensions determined from the activation-contractility models.

5 Conclusions and Recommendations

Due to the lack of direct implanted transducers, indirect monitoring of contractility based on activation-contractility modeling are only feasible solution, involving an index of activation, the observation of force-length and force-relationship and the activation-contractility modeling. Hence, efforts have been given on the real-time monitoring on the intensity of skeletal muscles' contraction (which is actually the monitoring of indexes for activation of muscle contraction), especially on the non-invasive approaches. A brief overview of both classical and up-to-date new technologies has been presented.

Up to date, the classic sEMG and MMG have not been applied in quantitative studies in dynamic motions, although they are used as an index of muscle activation in isometric or static condition. sEMG and MMG encountered significant noises problems especially in field activities. Imaging methods such as ultrasound scanning, despite their inconvenience for out-laboratory or in-field applications, have inspired a variety of other novel technologies on monitoring muscle contraction, among which continuous anthropometric measurement based on soft sensors appears to be effective and have been repeatedly understood as another index of activation and introduced in strain-contractility modeling. Anthropometric measurement based on fabric strain gauges also show great potentials for dynamic conditions such as in-field training. Applying such anthropometric techniques requires better understanding of strain-activation mechanism. Moreover, status parameters of contraction status, i.e., muscle length and shortening velocity, need to be obtained experimentally for independent monitoring of contractility,

In activation-contractility modeling, the effects of status parameters on muscle contraction, the force-length and force-velocity properties should be determined. For the force-length relationship, it is not recommended to scale an ideal biophysical parameterized force-length relationship for sarcomere to fit the target skeletal muscle, due to too many influencing factors difficult to reflect. To obtain the force-length relationship experimentally, there is a challenge for maintaining the 'intention of muscle contraction' (activation). Meanwhile, the difference between the torque-joint angle relationship and the real force-length relationship of skeletal muscle obtained in isometric tests shall also be noted. With regard to the force-velocity relationship, one should be aware is that the 'force' is either an index of load/resistance or contractility deducted inversely from devices such as force plates while the 'velocity' is not exactly the shortening velocity but a speed of the motion, only serves as representative of skeletal muscles' shortening velocity only. Literatures show that both the activation and muscle length impact the observed force-velocity relationship. While the force-velocity relationship of isolated muscles has been known to be hyperbolic, the multi-joint functional tasks typically reveal strong and approximately a linear force-velocity relationship, though biological mechanisms remain elusive.

The accuracy of a Hill type model on predicting muscle characteristics and contractility relies on the bio-signals as indexes of activations, the force-length and force-velocity relationship. To construct a Hill-type biomechanical model for in-motion monitoring of contractility, it is desirable to determine the core elements of a Hill type muscle through isolated designed tasks and selected conditions. However, since the model of a single sarcomere and that of a muscle may reveal huge difference due to the assumption of averaged shortening of sarcomeres and the recruitment patters of slower and faster motor units, as well as impedance caused by connective tissue and bones, it is recommended to give comprehensive consideration of the effect of specified length and velocity while focusing on the activation-contractility correlation, instead of product of the obtained factors as in the classical Hill's model.

Furthermore, the status of contraction, i.e, tension, muscle length and shortening speed are represented by joint torque, joint position (or joint angle) and joint angular velocity, respectively in the human musculoskeletal system, based on angle-to-length and angular velocity-to-shortening speed transform functions either with previously reported anatomical parameters or inversely derived from experiments. Hence, direct measurement of joint angles is not only essential for determination of force-length and force-velocity relationships but also for independent monitoring of contractility in-motion. A review of joint angle measurement technologies is presented, indicating that for in-motion conditions, soft goniometers are potentially better choices compared to camera-based motion capture system that is limited in laboratory and IMUS hindering normal movements. Further research is required to enhance accuracy, reliability of soft goniometers. If successful, an activation-contractility solution for motions can be completed and utilized as the effective monitoring, diagnosis and rehabilitation tools.

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