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Review

Motion analysis for the evaluation of muscle overactivity: A point of view

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ABSTRACT

Muscle overactivity is a general term for pathological increases in muscle activity such as spasticity. It is caused by damage to the central nervous system at the cortical, subcortical or spinal levels, leading to an upper motor neuron syndrome. In routine clinical practice, muscle overactivity, which induces abnormal muscle tone, is usually evaluated by using the Modified Ashworth Scale or the Tardieu Scale. However, both of these scales involve testing in passive conditions that do not always reflect muscle activity during dynamic tasks such as gait or reaching. To determine appropriate treatment strategies, muscle overactivity should be evaluated by using objective measures in dynamic conditions. Instrumental motion analysis systems that include 3-D motion analysis and electromyography are very useful for this purpose. The method can be used to identify patterns of abnormal muscle activity that can be related to abnormal kinematic patterns. It allows for objective and accurate assessment of the effects of treatments to reduce muscle overactivity on the movement to be improved. The aim of this point-of-view article is to describe the utility of instrumental motion analysis and to outline both its numerous advantages in evaluating muscle overactivity and to present the current limitations for its use (e.g., cost, the need for an engineer, errors relating to marker placement and cross talk between electromyography sensors).

1. The challenge of treating muscle overactivity

Muscle overactivity is caused by damage to the central nervous system at the cortical, subcortical or spinal levels, leading to an upper motor neuron (UMN) syndrome. Muscle overactivity includes spasticity (abnormal velocity dependent stretch reflex), dystonia (abnormal muscle contraction occurring during a voluntary movement) and abnormal co-contraction (abnormal simultaneous activation of agonist and antagonist muscles). Although these symptoms are each the result of different neuro-pathophysiological processes, they tend to be generically termed "muscle overactivity", which is the term we have chosen to use in this point-of-view article.

Muscle overactivity is widely believed to impair movement and function and thus is the focus of much medical treatment. A major challenge in treating muscle overactivity is the accurate evaluation of treatment effects. Although reduced muscle overactivity may lead to improvements in joint kinematics, it does not always lead to improvements in functional outcomes such as gait speed. In routine clinical practice, muscle overactivity associated with abnormal muscle tone is usually evaluated by using high-speed passive muscle stretching. The most commonly used tests are the Modified Ashworth Scale (MAS) and the Tardieu scale. These tests involve high-speed passive muscle stretching. Although intra- and inter-rater reliability have been found satisfactory [1], these tests provide only a measure of resistance to passive movement [2], which may not reflect muscle activity during dynamic tasks such as gait or reaching. The other main limitation of such scales is that they do not discriminate between different forms of passive resistance, in particular soft-tissue contracture, dystonia and spasticity.

Instrumental motion analysis (IMA) systems associating 3-D motion analysis, force platforms and electromyography (EMG) simultaneously record kinematic, kinetic and EMG parameters. They can be used to evaluate muscle overactivity during functional tasks and thus identify the mechanisms underlying a particular movement pattern. Following treatment, changes in various components of gait can be evaluated to help gain an understanding of why the patient feels an improvement: for example, an increase in knee flexion velocity during swing may not improve gait speed because of other impairments, but the patient may nevertheless

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feel less "stiff" and thus walk with more confidence and ease. Finally, IMA systems are also increasingly being used to study gait disorders in patients with UMN syndromes, both in clinical practice and research, although its use is less common for upper-limb movement assessment.

Although simpler methods do exist to objectively measure kinematic and spatiotemporal movement parameters, such as electrogoniometers, video recording, gait mats etc., they do not provide the same comprehensive level of evaluation as 3-D optoelectronic systems coupled with force platforms and EMG, which provide a complete, simultaneous analysis of kinematics (e.g., translation and angular displacement of body segments relative to each other), kinetics (e.g., internal and external moments of force) and patterns of muscle activation (EMG).

The aim of this point-of-view article is to summarise the principal findings of studies of the effects of treatments for muscle overactivity based on IMA systems. We aim to show how IMA systems can improve an understanding of the effect of muscle overactivity on motion and its utility in evaluating treatment effectiveness. For completeness, current limitations of IMA systems are also discussed.

2. IMA systems

The most common IMA systems are composed of optoelectronic cameras, force platforms, and EMG electrodes, all housed within a specifically designed motion analysis laboratory.

3-D motion analysis involves retroreflective passive markers (that reflect light emitted by the cameras) (Fig. 1a) or active markers (Fig. 1b) that are fixed to surface landmarks on the patient's body. The cameras record the 3-D trajectory of the markers, and specialised software then calculates kinematic parameters such as joint angles, angular velocities, translation of body parts etc. (Fig. 2). Spatiotemporal parameters can also be calculated, including speed, cadence, stride length and duration of the different phases of the gait cycle (single support, double support and swing phases) (Fig. 3), and, for upper-limb movements, metrics such as hand velocity, trajectory smoothness and curvature can be determined.

Force-platforms (Fig. 1b) embedded in the floor are used to quantify forces (kinetics) produced during the support phase of gait. An inverse dynamics model is used to calculate external moments (product of the force vector times the distance of the point of application of the force from the joint centre) and internal moments (caused by muscle contractions and ligamentous stability) (Fig. 2).

EMG (Fig. 2) is used to record muscle activity and usually involves recording of superficial muscles using surface electrodes. For the analysis of deeper muscles, fine-wire electrodes are required. EMG is used to study the temporal pattern of activation of the muscles and to determine whether the pattern is "normal" or "abnormal". Of note, the amplitude of the EMG signal does not reflect muscle strength.

The analysis of these simultaneously recorded parameters allows the clinician to:

- understand and explain movement anomalies;
- propose appropriate treatment;
- evaluate the effect of treatments on movement (gait or upper-limb movements).

3. Indications for the use of IMA to evaluate muscle overactivity

IMA is indicated in the case of muscle overactivity that has been clinically determined to be troublesome and for which the clinician

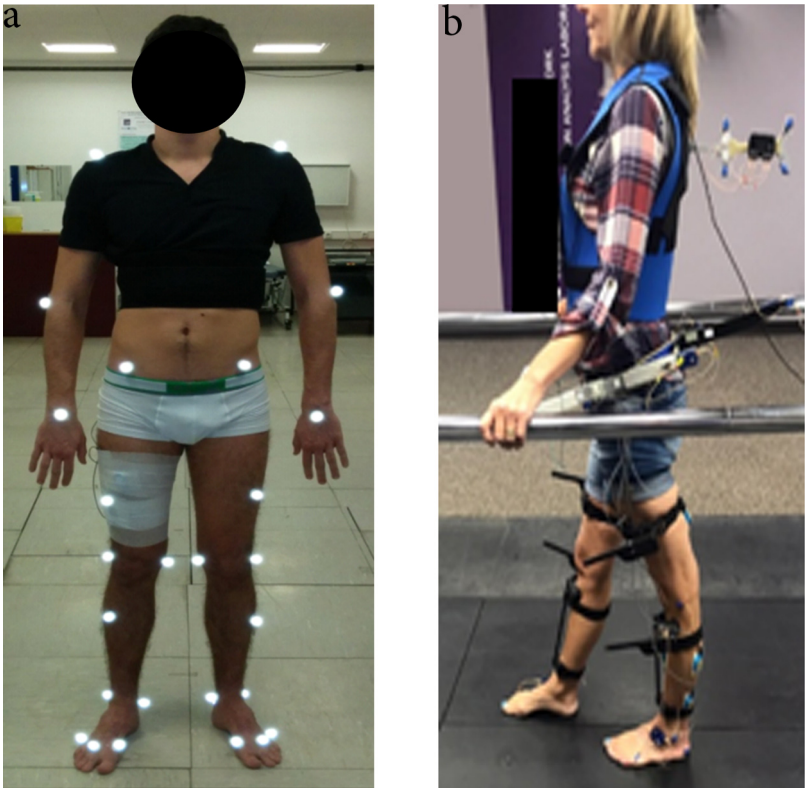

Fig. 1. a: patient wearing passive reflective markers positioned according to the Helen Hayes marker protocol; b: patient equipped with an active marker system (CODA Motion) and electromyography (EMG) stepping on force platforms for the simultaneous recording of kinematic, kinetic and dynamic EMG parameters during walking.

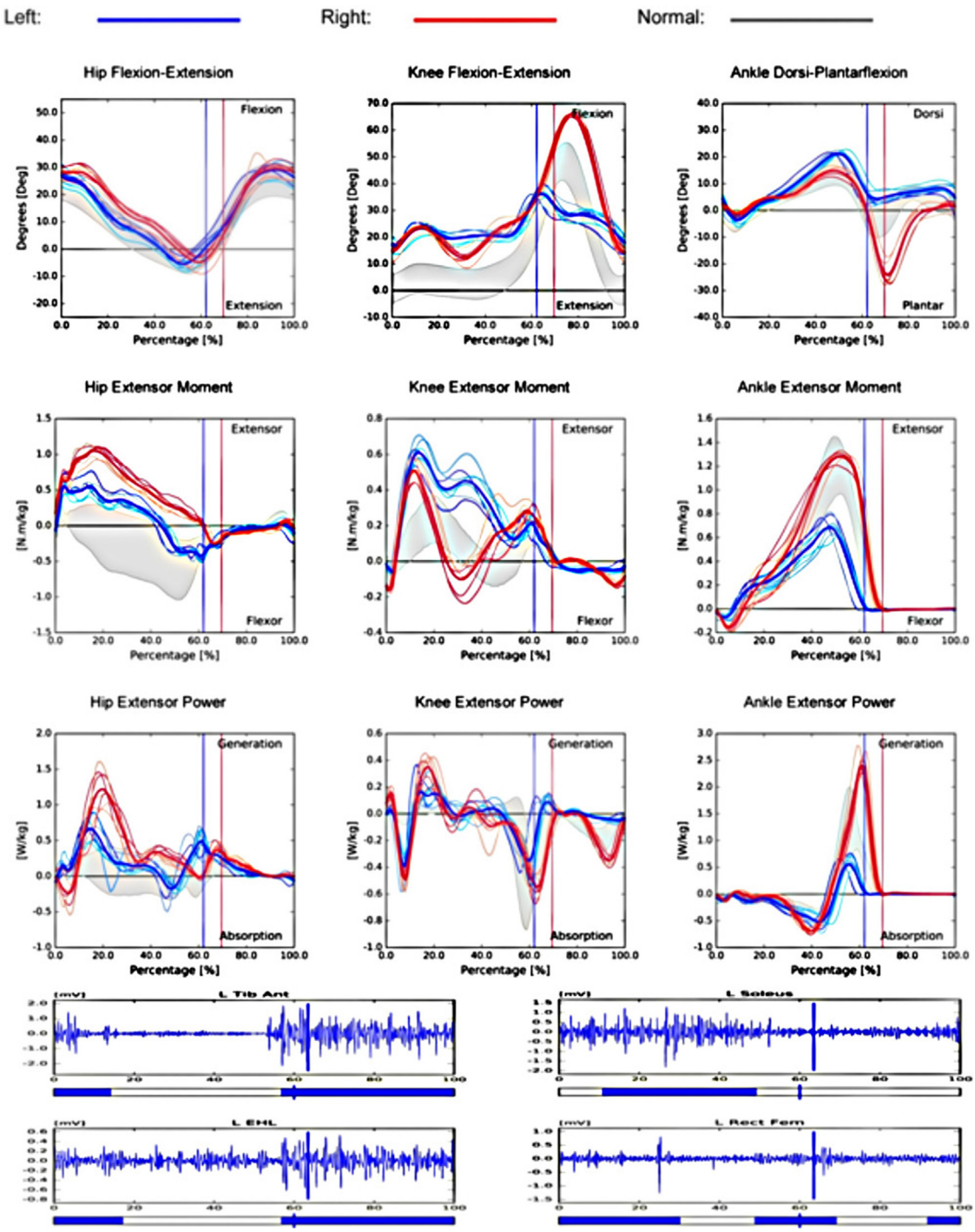

Fig. 2. 3-D kinematic and kinetic data summary sample with time-synchronised dynamic EMG obtained from a patient with left hemiparesis. The blue curves indicate mean values for the left, hemiparetic limb and the red curves indicate the right limb. The vertical lines indicate toe off for the left (blue) and right (red) limbs. The thick grey line indicates normal mean values with their SD values. For the EMG, only data for the left, hemiparetic limb are shown. The blue bars below indicate the periods in which each muscle is normally active. L Tib Ant: left tibialis anterior muscle; L Soleus: left soleus muscle; L EHL: left extensor hallucis longus; L Rect Fem: left rectus femoris.

wishes to provide interventional treatment, either pharmacological or surgical. IMA, combined with a thorough clinical assessment (particularly of spasticity, muscle strength and passive range of motion) has revealed certain underlying dysfunctions that cause pathological gait in patients with hemiparesis. EMG analysis combined with a clinical assessment can reveal the causes of abnormal kinematic parameters. For example, studies involving IMA have shown the following:

- decreased hip extension during stance phase may be due to weakness of the hip extensors or to overactivity or shortening of the hip flexors;
- decreased hip flexion during swing phase may be due to weakness of the hip flexors or to overactivity of the hip extensors (gluteus maximus or the hamstring muscles). This condition is frequently associated with a lateral tilt of the pelvis due to compensation by trunk muscles such as quadratus lumborum;

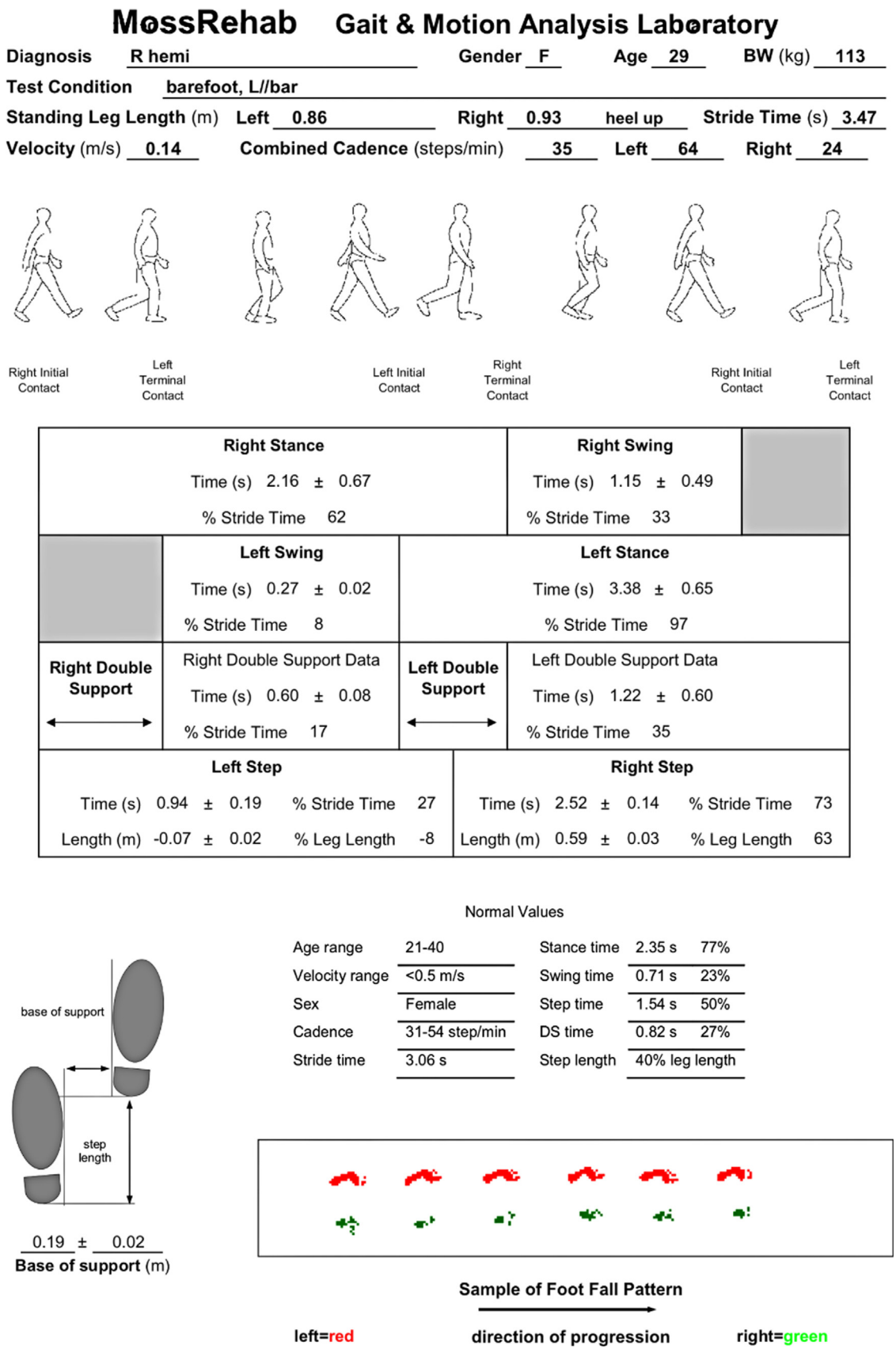

Fig. 3. Sample of spatiotemporal data from a patient with left hemiparesis with sex-, age- and velocity-matched normative data. The durations of the different phases of the gait cycle for each limb are provided, along with normative values.

- decreased peak knee flexion in swing phase associated with a dumpling in the angular displacement curve can be due to several mechanisms such as weakness of the hip flexors, a reduced plantar flexion moment in late stance or inappropriate activity of part of the quadriceps muscle, particularly the rectus femoris (RF) during the swing phase of the gait cycle [3]. Inappropriate activation of the RF, in particular during swing phase, restricts peak knee flexion during swing [3–5];
- decreased ankle dorsiflexion in stance phase is commonly due to overactivity of the gastrocnemius or soleus or a shortening of these muscles and is frequently associated with an increase in knee extension during weight bearing;
- decreased ankle dorsiflexion during swing or at heel strike may be due to weakness of the dorsiflexor muscles, particularly the tibialis anterior, or to inappropriate activity of the triceps surae (TS) [6,7].

Use of IMA has also confirmed clinically well-known spatio-temporal alterations, such as decreased gait speed, step length and cadence of the paretic lower limb, and demonstrated that the time spent in single support phase and in double contact are increased for the non-paretic lower limb [8,9].

IMA is useful for assessing lower-limb muscle coactivation and its relationship with muscle overactivity and gait performance. For example, in a study of 23 patients with hereditary spastic paresis and 23 controls, the coactivity index of both the knee and ankle muscles was increased throughout the gait cycle and during the sub-phases of gait in patients versus controls. The authors also showed that both knee and ankle coactivation indexes were positively correlated with spasticity (Ashworth score) [10].

Gross et al. [9] investigated the relationship between muscle coactivation and joint kinematics in patients with hemiplegic cerebral palsy. The authors calculated a coactivation index of knee and ankle agonist/antagonist muscles (RF/semi-tendinous, vastus medialis/semi-tendinous and tibialis anterior/soleus muscles) at 3 different gait speeds in 12 patients with hemiplegic cerebral palsy and 12 controls. In the controls, a higher coactivation index was associated with reduced joint excursion for 7 of the 14 linear models (hip flexion, knee loading, knee extension in stance, knee flexion in swing, ankle plantarflexion from initial contact to foot flat, and ankle dorsiflexion in stance and in swing). Conversely, in involved limbs of children with cerebral palsy, higher coactivation indexes were associated with reduced knee loading, ankle plantarflexion excursion at push off and ankle dorsiflexion in swing and increased hip excursion.

Muscle synergies have also been studied in patients with cerebral palsy by using IMA. Steele et al. evaluated muscle synergies during gait in 84 healthy individuals and 549 patients with cerebral palsy. Synergies were calculated by using non-negative matrix factorization on surface EMG data. As compared with healthy controls, patients with cerebral palsy used fewer synergies, which suggests a simplified control strategy during gait [11].

In 38 patients with multiple sclerosis, IMA showed that spasticity alters the gait pattern by reducing gait speed, stride length and swing phase duration and increasing double support time. It also revealed specific alterations in kinematic parameters and muscle activation, such as reduced hip and knee flexion–extension range of motion and abnormal activation of the RF muscle [12].

As demonstrated by the studies described above, IMA facilitates analysis of the interactions between muscle overactivity, muscle coactivation or muscle synergies, and kinematic alterations. We believe that such comprehensive analysis is essential to fully understand the movement disorder of each individual before planning treatment to reduce muscle overactivity.

4. Use of IMA to evaluate the effect of treatments on reducing overactivity of lower-limb muscles and improve gait

Probably the most widely used treatment to reduce muscle overactivity is botulinum toxin type A (BoNT-A) injection. BoNT-A works presynaptically to inhibit acetylcholine release, decreasing output at the neuromuscular junction [13] and hence reduce muscle overactivity.

5. Single-session BoNT-A injection

3-D motion analysis of gait at a comfortable speed has been used to demonstrate improvements in gait kinematics after BoNT-A injection: BoNT-A injection in the RF muscle was found to increase peak knee flexion during the swing phase of the gait cycle by 5° to 9°, reducing stiff knee gait [4,5], and also increase peak hip flexion by 3° to 5°. Similarly, BoNT-A injection in the TS increased peak ankle dorsiflexion in swing phase by 1.5° but not significantly [6]. This finding is useful to orientate further research and guide clinical practice.

Although 3-D motion analysis has shown improvements in kinematic parameters after BoNT-A injection, the effects on spatiotemporal parameters are inconsistent. One large placebo-controlled study ($n = 234$) did not find any improvement in gait velocity after BoNT-A injection in the plantarflexor muscles (despite a reduction in spasticity) [6], but several other studies found improvements in gait velocity ($n = 45$ and $n = 12$) [14,15]. Similarly, after BoNT-A injection in the RF muscle that led to an improvement in stiff knee gait, only one study found increased gait velocity [16], but several others found no change in gait speed [4,5]. It therefore seems unlikely that the changes in kinematic parameters were due to any increase in gait speed.

As can be seen from the above data, the recorded improvements in kinematic parameters do not always correspond to quantifiable improvements in the functional tests commonly used in routine practice. For example, a randomised controlled study without IMA showed that for 16 patients with chronic stroke who received only one BoNT-A injection performed in accordance with patients' needs (e.g., in gluteus magnus, hamstrings, RF, TS, toe flexors digitorum longus, flexors halluxis longus, etc.), no improvements were found in performance of the 10-m walk test, Timed Up and Go test, 6-min walk test or Stair Test at 1 month later [17].

Moreover, all kinematic results are presented in the sagittal plane, mainly because the data obtained by motion analysis systems are more reliable versus data from the frontal and transverse plans [18,19]. Nevertheless, a few studies presented kinematics in frontal and transversal plans after BoNT-A injection in patients with neurological pathologies. Papadonikolakis et al. described changes in the sagittal plane after BoNT-A injection in the lower-limb muscles of patients with cerebral palsy, although they analyzed hip kinematics in the frontal plane (presented in the Table but not cited) [20]. Sutherland et al. assessed the effects of BoNT-A injection into the gastrocnemius muscle on gait of patients with cerebral palsy. In addition to the increase in dorsiflexion ankle peak (+8.5° in stance, +5° in swing), the authors reported a decrease of the external rotation of the foot in swing phase [21].

6. Multi-site BoNT-A injection

Patients with UMN syndromes rarely have overactivity of a single muscle. Therefore, treatment usually involves injecting several muscles or muscle groups. Few studies using IMA have evaluated the effect of multi-site injection of BoNT-A. Caty et al. (2008) assessed the effect of a BoNT-A injection in the RF, semi-tendinous and TS muscles [22]. Twenty patients received

injections with the aim of improving their stiff knee gait; IMA 2 months later revealed a 5° increase in peak knee flexion during the swing phase of the gait cycle. External mechanical work was also reduced, which indicates that treatment involving multi-site BoNT-A injection decreases the energy cost during gait [22].

Marchiori et al. assessed the effect of multisite BoNT-A injection in adults with cerebral palsy by using the Gait Deviation Index. Post BoNT-A injection, stride length, peak hip flexion and peak knee flexion during swing phase were significantly increased, but the Gait Deviation Index was not significantly modified [23].

7. Repeated BoNT-A injection

The effect of BoNT-A injection usually lasts for about 3 months, so injections are repeated at about every 4 months. Regardless, we found only one study that evaluated the effect of consecutive multifocal BoNT-A injection. This multicentre, prospective randomized, placebo-controlled study of 352 adults with chronic hemiparesis evaluated BoNT-A injection into several muscles of the lower limb that were repeated every 3 months over 2 years, according to the individual needs of patients [24]. Kinematic analysis showed an increase of 12° in knee range of motion during swing and spatiotemporal analysis showed a 20% to 24% improvement in gait velocity. The authors found a 9.4% to 13.8% improvement in stride length and a 7.2% to 9.7% improvement in cadence (depending on whether the patient walked at a comfortable or fast speed, barefoot or with shoes). Moreover, these improvements were more marked after each injection session. These data suggest that adaptation to reduced muscle overactivity requires time, although more studies are needed both to confirm these findings and to further guide treatment practice.

8. Other treatments to reduce muscle overactivity

IMA has been used to evaluate other muscle-overactivity reduction treatments. One study evaluated the effect of functional tendon surgery for equinovarus foot caused by muscle overactivity and contracture in 177 patients with chronic stroke. The IMA was performed before surgery and repeated 1 year later; the analysis revealed small improvements in kinematic parameters of the ankle joint as well as spatiotemporal gait parameters [25].

9. Use of instrumental gait analysis for preliminary determination of the effect of BoNT-A injection

Although agents such as BoNT-A effectively reduce muscle overactivity [26], the effect on kinematic parameters is, as shown above, often modest probably because only mean or median responses are reported for studies. Thus, the effect may be negligible in some patients and large in others. Such reporting may also explain the mixed results regarding improvements in spatiotemporal gait parameters and functional assessments. In our opinion, research is needed to refine treatment indications and determine the characteristics of patients who would most benefit from muscle activity reduction. One study has been undertaken in this area; results indicated that the degree of improvement in peak knee flexion 1 month after BoNT-A injection in the RF was positively correlated with the patient's pre-injection capacity to increase peak knee flexion in a fast gait speed condition [27], which suggests that the behaviour of the knee joint during fast walking speed should be assessed before injection. More studies are needed to determine whether such findings can be applied to all patients with muscle overactivity and what, if any, other parameters may be of prognostic value regarding the effect of overactivity reduction treatment.

10. Musculoskeletal models

In recent years, musculoskeletal models such as SIMM® [28] and open SIMM® [29] have been used to dynamically simulate movement and assess muscle length and force changes produced during gait in patients with UMN disorders. The generic, open-access SIMM® software generates a model consisting of 13 rigid segments (pelvis, and femur, tibia, patella, heel, foot and toes of both limbs) with 17 degrees of freedom. Each lower limb consists of 43 musculotendinous complexes. Each complex is defined by its origin and insertion and if necessary, by defined via points in order to specify the muscle trajectory as much as possible. This approach is particularly interesting for predicting the effects of different types of treatment, whether pharmacological or surgical: patient data can be input into the model and simulations of the effects of different "treatments" can be performed.

Such models can also be used to gain a better understanding of the mechanisms behind gait abnormalities and behind treatment effectiveness or lack of effectiveness. For example, SIMM® modeling software was used to evaluate the effect of BoNT-A injection in the RF on the length of the RF musculotendinous complex in 10 patients with hemiparesis and stiff knee gait. One month after the injection, maximal RF length and maximal lengthening velocity were increased [30]. These data suggest that BoNT-A injection could also modify muscle spindle sensitivity, which supports previous results [31,32].

Open SIMM® has also been used to predict surgical outcomes in patients with cerebral palsy. Simulation was used to predict the effects and determine the indications for RF transfer surgery on knee flexion in individuals with cerebral palsy [33,34].

These results support the idea of possibly predicting the gait pattern of a patient before treatment, whether pharmacological or surgical, in the future.

11. Intersegmental coordination analysis

Inter-segmental coordination analysis involves the correlation of a number of parameters to describe spatiotemporal organization between segments and to identify gait patterns in specific planes. This analysis contrasts with focal gait analysis that describes individual joints. A common method to analyse intersegmental coordination is the Continuous Relative Phase (CRP) metric [35–37], which continuously correlates angle positions and velocities of 2 segments during the different movement phases. The CRP between 2 segments is defined as the difference between the 2 phase angles at any point of the gait. A CRP of 0° indicates that the segments move in-phase (as in a stiff leg with no movement between the thigh and the shank). A positive CRP indicates that the thigh "leads" the shank. As the CRP increases, the thigh and shank are increasingly out-of-phase until a CRP of 180°, which indicates anti-phase coupling [35–37] (Fig. 4). This method has been used to evaluate intersegmental coordination in patients with stroke and showed that thigh-shank coordination ($CRP_{Thigh\text{-}Shank}$) during swing was reduced in the paretic limb before BoNT-A injection in the RF and improved significantly after injection [35]. Moreover, in the non-paretic limb, $CRP_{Thigh\text{-}Shank}$ tended to be increased before injection and decreased significantly afterwards (Fig. 5). Hence, this method is useful to assess both inter-joint coordination within the affected lower limb and compensatory strategies in the less-affected limb.

12. Recommendations and limitations for the use of 3-D motion analysis to evaluate gait and for data interpretation

The principal limitation of IMA systems is the cost, not only the initial purchase but also over the long term, because a technician or engineer usually operates the system and processes the data for

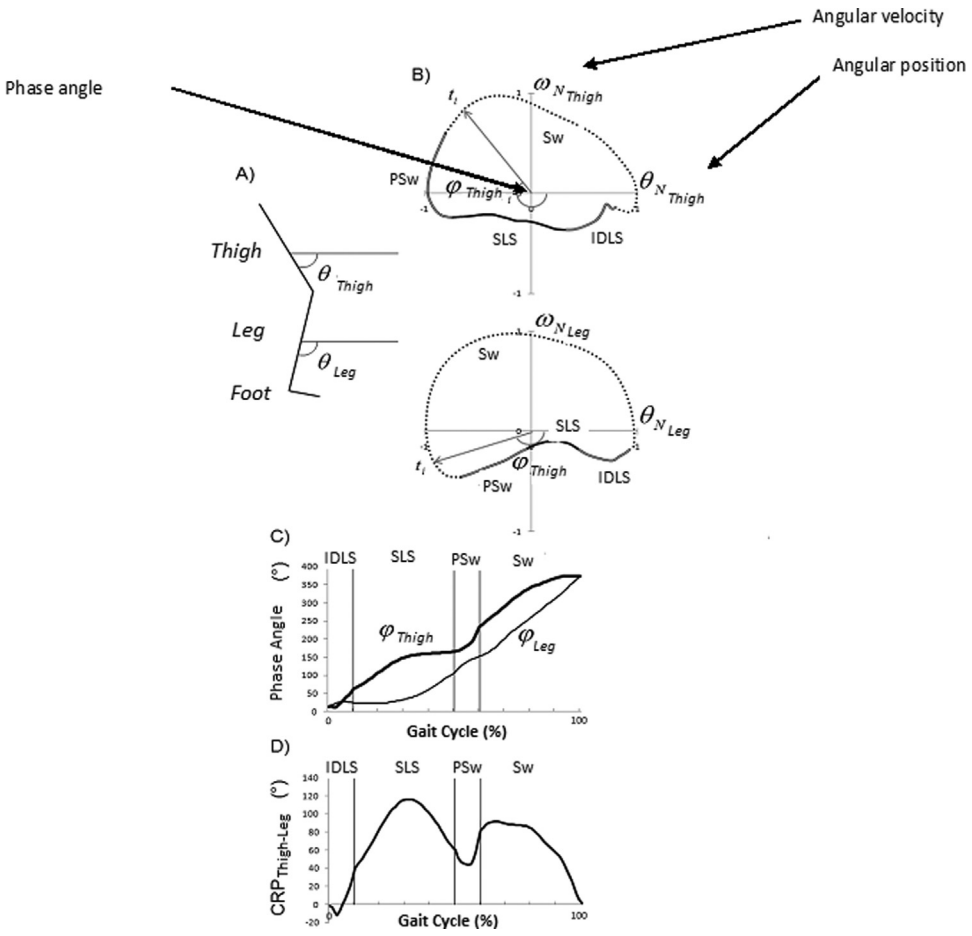

Fig. 4. Variation of the phase angle of the thigh and the shank during a gait cycle. SLS, single limb stance; IDLS, initial double limb stance; PSw, Pre swing; Sw, Swing, From Hutin et al., 2010.

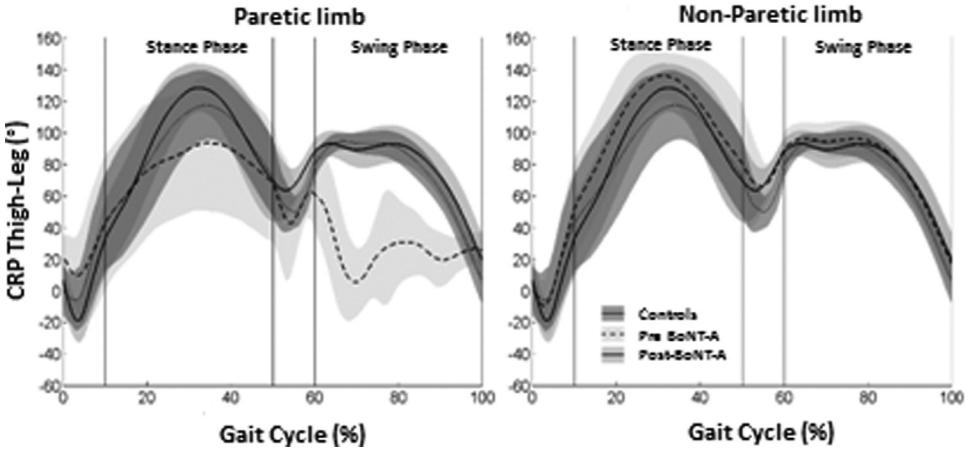

Fig. 5. Continuous Relative Phase data for the paretic (left graph) and non-paretic limbs (right graph) before (hatched line) and after (solid line) BoNT-A injection compared to normative data (from healthy subjects) in dark grey. From Hutin et al., 2010.

clinicians to then analyse and interpret. Regardless, we argue that such systems are indispensable for managing movement disorders in patients with UMN syndromes. Use of IMA facilitates accurate diagnoses, effective treatments, effective use of clinical time and resources and reduced pain and costs (to both patients and healthcare providers) associated with unnecessary or inappropriate treatments. In the future, with new developments in technology, less costly, more user-friendly systems will likely become available [38].

Although 3-D motion analysis provides objective movement evaluation, the use of the system requires training, and care must be taken to ensure that the data produced are accurate. For example, errors can arise due to skin movement during measurement or re-positioning of markers and EMG electrodes between

recording sessions. To minimise the risk of such errors, a skilled technician or clinician must position markers and electrodes at each session.

Also, appropriate, validated protocols must be used for positioning markers and EMG electrodes. If possible, International Biomechanical Society guidelines should be followed, such as the Helen Hayes protocol for gait analysis (Fig. 1). International standardisation of protocols allows for easier comparison of the results of different studies.

In addition, gait recorded under laboratory conditions in which the patient is in a state of semi-undress and covered in adhesive markers is not entirely a natural, real-life scenario, which leads to an increase in the variability of gait parameters [39]. However, a study by our group demonstrated that after repeated sessions, patients become accustomed to the procedure and this renders their gait less variable [40]. Therefore, we recommend performing a "familiarisation" session, when possible, before any data collection for therapeutic analysis.

Regarding EMG limitations, the surface EMG does not allow for analyzing the activation pattern of the deep muscles. The EMG record may include unwanted signals arising from neighboring muscle (cross talk), which implies greater activity. Tissue displacement accompanying a muscle contraction or related to the mass of the adipose tissue can interfere with the signal. In addition, floor impact during walking can generate signal artefacts. To prevent these limitations, the skin could be cleaned at the electrode location by shaving and dirt removed with alcohol for low impedance.

Regarding limitations of kinetics parameter interpretation, the foot must be fully on the force platform to analyze kinetic gait parameters. We recommend recording a minimum of 3 trials to obtain a precise measure of force platform [41]. Considering forces in 3-D obtained with the force platform, the vertical ground reaction force is considered the most reliable measure of kinetic data [42,43].

When interpreting results, one must be aware that there is already a natural variability in all human movements. Therefore, any change in a patient's movement parameters after a particular treatment must consider this natural variability. One study determined minimal detectable change (MDC 95%) values for kinematics, ground reaction force, spatial and temporal variables collected during treadmill walking post-stroke [44]. Our team recently determined MDC 95% values for the same parameters during overground walking in 26 patients with stroke. The recordings were performed with participants barefoot at a spontaneous walking speed without orthoses. All patients underwent 3 instrumental motion analyses at 7-day intervals. The Helen Hayes protocol was used (Table 1) [40]. The MDC was calculated by the following equation: $MDC = SEM \times 2.056 \times \sqrt{2}$. The value 2.056 corresponds to the Student t distribution with a 95% confidence interval for the study sample size ($n = 26$) [40]. The SEM and MDC are expressed in raw units and not in percentages to facilitate future comparisons with other studies. However, the mean SEM and MDC values for the spatiotemporal parameters are presented as percentages to allow for comparison of parameters with different units [45]. The MDC% was calculated with the following formula: $MDC\% = \left(\frac{MDC}{X}\right) \times 100$, where X is the mean value of the parameter compared between 2 sessions (V1/V2, V1/V3 or V2/V3) [46]. The SEM% was calculated in the same way. The MDC% for most gait parameters was lowest between the second and third visits, which suggests that patients should attend a familiarization session before the actual evaluations. However, such evaluation is expensive and time-consuming. Nevertheless, this process would ensure that changes measured are related to the treatment and are not an effect of the repeated evaluations. It would be interesting if all studies mentioned in this article would have done this analysis to discuss more serenely all previous results.

Table 1
Minimal detectable change (MDC) of kinematic and spatiotemporal parameters of the paretic lower limb during overground gait [35].

		MDC 95%
Kinematic parameters		
Hip maximum angle	SwP	9.0°
	StP	7.3°
Hip minimum angle	SwP	7.6°
	StP	6.5°
Hip RoM	SwP	4.7°
	StP	4.0°
Knee maximum angle	SwP	6.5°
	StP	4.9°
Knee minimum angle	SwP	5.9°
	StP	5.5°
Knee RoM	SwP	6.4°
	StP	5.2°
Ankle maximum angle	SwP	5.5°
	StP	4.9°
Ankle minimum angle	SwP	5.79°
	StP	5.9°
Ankle RoM	SwP	2.0°
	StP	3.9°
Spatiotemporal parameters[a]		
Gait speed		14.6%
Stride length		11.9%
Cadence		8.6%
Step length		6.3%
Stance phase duration		3.6%
Swing phase duration		3.6%
Step width		2.45%

RoM: range of motion; SwP: swing phase; StP: stance phase.
[a] For the spatiotemporal parameters, MDC values are presented as percentages to rationalize and compare the different units.

13. Use of IMA to evaluate the effect of treatments to reduce overactivity of upper-limb muscles

Fewer studies have evaluated the effects of muscle overactivity on upper-limb movement than lower-limb movement. This situation is due to several factors: first, upper-limb musculoskeletal models are more complex than lower-limb models; second, there is no "typical" task to evaluate upper-limb movement parameters as there is gait for the lower limbs; and third, surface EMG measurements are particularly difficult to record in the upper limb because of cross-talk between muscles; it is difficult ensure that recorded activity actually corresponds to the muscle over which the electrode is positioned.

Despite these challenges, a few studies have evaluated the utility of IMA to evaluate reaching and grasping movements in patients with upper-limb muscle overactivity.

IMA has demonstrated the typical characteristics of upper-limb motor control in stroke. There is typically a reduction in velocity of the hand during reaching movements, with several velocity peaks [47,48]. The trajectory of the hand is also more curved and less smooth [49,50]. These kinematic abnormalities mostly result from muscle co-contractions [51] and a loss of selective movement. Shoulder–elbow joint coordination is also altered [48] and patients often use compensatory strategies involving trunk motion to achieve their goal [51].

With regard to grasping, the aperture of the hand is reduced during the approach to grasp an object. The reduction in the activation of the finger extensor muscles [52–54] and in coordination between the extensor and flexor muscles leads to highly variable patterns of hand opening, which are often inefficient [55].

Musampa et al. used IMA in hemiplegic patients to assess the double joint shoulder–elbow system in 10 healthy individuals and 11 patients with stroke and spasticity. EMG activity from 4 elbow and 2 shoulder muscles was recorded during quasi-static stretching of the elbow flexors and extensors and during slow voluntary movement throughout full range. Stretches and active movements were initiated from full elbow flexion or extension with the shoulder in 3 different initial positions. The results showed that stretch reflex thresholds of the elbow muscles were correlated with clinical measurements of spasticity. The stretch reflex threshold of the elbow flexor and extensor muscles were also within the biomechanical range of the joint and varied with changes in shoulder angle. The authors concluded that there was a spatial zone in which spasticity was present and another in which it was absent. The boundary between the spasticity and the no-spasticity zones depends on the state of reflex inter-joint interaction [56].

Mandon et al. used IMA to evaluate the combined effects of reaching distance and speed instruction on trunk and paretic upper-limb coordination in 14 patients with stroke and spasticity. The reaching movements were performed to 2 targets at 60% and 90% of arm's length, respectively, and at preferred and maximum velocity. The experiment was performed with the trunk free and the trunk restrained by a strap. All patients were able to increase their reaching velocity. However, in the trunk-free, faster speed condition, elbow extension velocity increased, but elbow extension amplitude decreased and trunk movement increased. In the trunk-restraint condition, elbow extension amplitude did not decrease with faster speeds. The authors concluded that faster speeds may encourage some patients to use compensatory strategies [57].

Chang et al. determined the most appropriate kinematic variable to quantify the effects of muscle overactivity on reaching in children with cerebral palsy (10 healthy individuals and 10 with cerebral palsy). Prolonged movement time and reduced smoothness both had a larger effect size than did other kinematic variables. As well, movement time in children with cerebral palsy was prolonged in comparison with healthy children. Moreover, the MAS score was significantly correlated with jerk, number of movement units and movement time. The number of movement units was the most sensitive kinematic variable to discriminate between healthy individuals and those with spasticity during reaching [58].

Robust evidence now supports that BoNT-A injection reduces resistance to passive movement, as measured by the MAS, and improves self-care ability for the affected hand and arm but does not improve arm and hand use. IMA would be useful to determine the effect of BoNT-A injection on active spastic upper-limb movement [59].

A recent case study by Sarcher et al. (2018) [60] described a novel method to detect deviations in the patterns of muscle activity in a child with unilateral spastic cerebral palsy. The method involved using data from healthy children to control for natural EMG variability. Using surface EMG of the biceps, triceps, brachioradialis, pronator teres, pronator quadratus, and brachialis muscles during elbow flexion-extension and pronation-supination tasks, the authors showed higher levels of muscle activity in the child with cerebral palsy than in the healthy children. The method also detected reductions in muscle activity after BoNT-A injection. Although this particular analysis involved EMG without kinematic analysis, future work could include an IMA system to provide a more detailed picture of upper-limb impairments.

Another study of upper-limb movement involving 3-D motion analysis evaluated the effect of BoNT-A injection on reaching and hand-to-mouth tasks in 10 children with hemiplegia [61]. Twelve weeks after treatment (injections in forearm muscles and 6 weeks of occupational therapy), elbow flexor muscle tone was reduced significantly but with no differences in the active range of movement recorded during the 3-D motion analysis. However, analysis of individual data showed that 2 weeks post-injection, active elbow extension had increased by $>15^{\circ}$ in 3 participants and forearm supination had increased by $>25^{\circ}$ in 6 participants during performance of the reach and hand-to-mouth tasks, respectively. These results suggest that individuals respond differently to BoNT-A injection.

The effect of BoNT-A injection in several upper-limb muscles, depending on the needs of each individual patient, on the kinematics of reaching [62] and grasping [63] movements was evaluated in one study of 15 adults with spastic hemiparesis. 3-D analysis showed significant improvements in the velocity and smoothness of the hand trajectory during reaching movements that was associated with an increased tendency to use the paretic arm [62]. Furthermore, video analysis of finger configuration during object grasping showed improvements in both the final direction of the approach and the hand grasping posture [63]. Analysis of individual data showed that the patients with the best potential for functional improvement were those with good proximal and moderate distal motor control.

The studies by Mackey et al. [61] and Bensmail et al. [62,63] both suggest that some patients may benefit more from BoNT-A injection than others. Further studies using 3-D motion analysis could be useful to determine whether certain characteristics could predict which patients are more likely to benefit from BoNT-A injection. Such characteristics may be the strength of agonist muscles or the presence of overactivity in other muscles or perhaps the capacity to increase reaching velocity; indeed, Roche et al. (2014) [27] found that the capacity to increase knee flexion during fast gait predicted the effect of BoNT-A injection in lower-limb muscles.

14. Recommendations and limitations regarding the use of IMA systems to evaluate upper-limb movements and for data interpretation

In humans, 3-D motion analysis of upper-limb movement is more complex than 3-D motion analysis of gait [64,65]. Gait is not only the main function of the lower limbs but is also relatively stereotyped across individuals, whereas there is not one predominant functional activity for the upper limb. Several functional tasks [66,67] have been suggested. Recently Alt Murphy et al. [67] described a standardized protocol for kinematic analysis of a drinking task applied in individuals with upper-limb impairments after stroke. For arm motion analysis to become routinely used for rehabilitation assessment and evaluation of treatment effects, a validated set of relevant tasks should be established.

3-D motion analysis of upper-limb movements is also associated with technical issues because the upper limb, and especially the shoulder joint, has a very large working range, as compared with the lower limb. This causes problems such as gimble lock and soft-tissue artefacts. To promote the standardization of 3-D motion analysis, the International Society of Biomechanics (ISB) has published recommendations regarding the definition of joint coordinate systems and rotation sequences for the trunk, shoulder, elbow, wrist and hand [68] that are based on an earlier ISB standard for reporting kinematic data [73]. A definition for the local axis system in each articulating segment or bone is provided for each joint while respecting the upper-limb anatomy and clinical interpretation of joint movements [68]. However, consensus on the entire protocol for 3-D motion analysis is still required by experts in the field in order to facilitate the comparison of results and encourage communication among researchers and clinicians.

15. Conclusions

We strongly believe, based on the studies presented in this point-of-view article, that IMA systems should be more widely used to evaluate movement in individuals with muscle overactivity. As demonstrated by the studies presented, IMA is useful to determine the impact of muscle overactivity on gait, which then allows for proposing suitable treatment. It can also be used to determine the effect of treatment to reduce overactivity on spatiotemporal, kinematic and kinetic parameters. Use of this method to evaluate upper-limb movements, although more complicated because of the anatomy and physiology of the upper limb, is also gaining attention for similar reasons. Clearly, standardized assessment tasks using standardized 3-D movement analysis protocols need to be validated to facilitate their use in routine clinical practice.

The major limitation of such systems is the need for technical expertise and specialist equipment; however, from research evidence as well as our own clinical experience, we firmly believe that some patients should benefit from IMA, particularly when there is some doubt at the time of the clinical examination or when treatment is ineffective.

Disclosure of interest

The authors declare that they have no competing interest.

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