



Quantitative magnetic resonance imaging towards clinical application in multiple sclerosis

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[†]Appendix 1.

Quantitative MRI provides biophysical measures of the microstructural integrity of the CNS, which can be compared across CNS regions, patients, and centres. In patients with multiple sclerosis, quantitative MRI techniques such as relaxometry, myelin imaging, magnetization transfer, diffusion MRI, quantitative susceptibility mapping, and perfusion MRI, complement conventional MRI techniques by providing insight into disease mechanisms. These include: (i) presence and extent of diffuse damage in CNS tissue outside lesions (normal-appearing tissue); (ii) heterogeneity of damage and repair in focal lesions; and (iii) specific damage to CNS tissue components. This review summarizes recent technical advances in quantitative MRI, existing pathological validation of quantitative MRI techniques, and emerging applications of quantitative MRI to patients with multiple sclerosis in both research and clinical settings. The current level of clinical maturity of each quantitative MRI technique, especially regarding its integration into clinical routine, is discussed. We aim to provide a better understanding of how quantitative MRI may help clinical practice by improving stratification of patients with multiple sclerosis, and assessment of disease progression, and evaluation of treatment response.

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Abbreviations: AD = axial diffusivity; DTI = diffusion tensor imaging; FA = fractional anisotropy; MD = mean diffusivity; MP2RAGE = magnetization prepared 2 rapid acquisition gradient echoes; MTR/sat = magnetization transfer ratio/saturation; MWF = myelin water fraction; QSM = quantitative susceptibility mapping; RD = radial diffusivity; T_1 -RT = T_1 relaxation time

Introduction

Conventional MRI provides invaluable information for the diagnosis, prognosis, and monitoring of the effectiveness of therapeutics in patients with multiple sclerosis.^{1,2} The term conventional MRI encompasses the methods used in clinical practice to describe pathology by relying on contrast changes in weighted images. These are images that predominantly, but not exclusively, reflect a biophysical contrast mechanism (e.g. T_1 - and T_2 -weighted scans). Using conventional MRI in the multiple sclerosis clinic, it is generally possible to identify the number, location, and activity of multiple sclerosis lesions, although the sensitivity to those characteristics generally varies depending on several technical factors.³ On the other end, conventional MRI is largely insensitive to the heterogeneity of focal multiple sclerosis lesions and to the pathology affecting CNS tissue outside multiple sclerosis lesions (normal-appearing white and grey matter). Furthermore, conventional MRI is unable to depict the level of damage within different CNS tissue components, such as myelin, axons, and glia.

Better quantification of the extent, type, spatial distribution, and evolution over time of CNS tissue damage in patients with multiple sclerosis could improve our understanding of disease mechanisms. It may also aid in stratification of disease burden, assessment of therapy response and evaluation of subclinical disease progression.

Quantitative MRI can potentially address these needs by providing more sensitive measures of multiple sclerosis pathology and more specific information regarding which tissue component has been damaged (Fig. 1). Unlike conventional MRI, which acquires datasets that have a mixture of weightings and therefore cannot be resolved into a quantitative map, quantitative MRI relies exclusively on acquisitions that can then be used to disentangle the source of signal variations. Moreover, through computational

or mathematical modelling, this approach can provide quantitative maps where intensities have physical units.⁴ Thus, quantitative MRI techniques are superior to conventional MRI regarding their sensitivity to subtle alterations within lesions and normal-appearing tissue⁴ as well as their increased specificity relating to the damage of different tissue components of the CNS (e.g. myelin, axons, glia, iron and blood flow/volume).

Nonetheless, quantitative MRI is not currently used in clinical practice, primarily because it has not reached 'clinical maturity'. A method can be considered 'clinically mature' when it can be run on most up-to-date clinical scanners without the need for additional sequence development, there is available and validated software able to process the data and provide the user with the desired quantitative maps, and cut-off values of pathology assessed with that method have been established.

In this review, we summarize: (i) the information that can, and cannot, be provided by conventional MRI; (ii) the contribution of quantitative MRI to our understanding of multiple sclerosis pathology in the brain and spinal cord; (iii) the relationship between quantitative MRI features and clinical outcome and the potential role of quantitative MRI in improving the prediction of disability, especially motor and cognitive deficits; and (iv) the clinical maturation stage of the various quantitative MRI techniques.

Quantitative MRI and multiple sclerosis neuropathology

When radiographically investigating multiple sclerosis, conventional MRI provides the following measures: (i) number, volume, and location of focal T_2 -weighted hyperintense lesions; (ii) number of contrast-enhancing T_1 -weighted lesions (CEL); (iii) number and

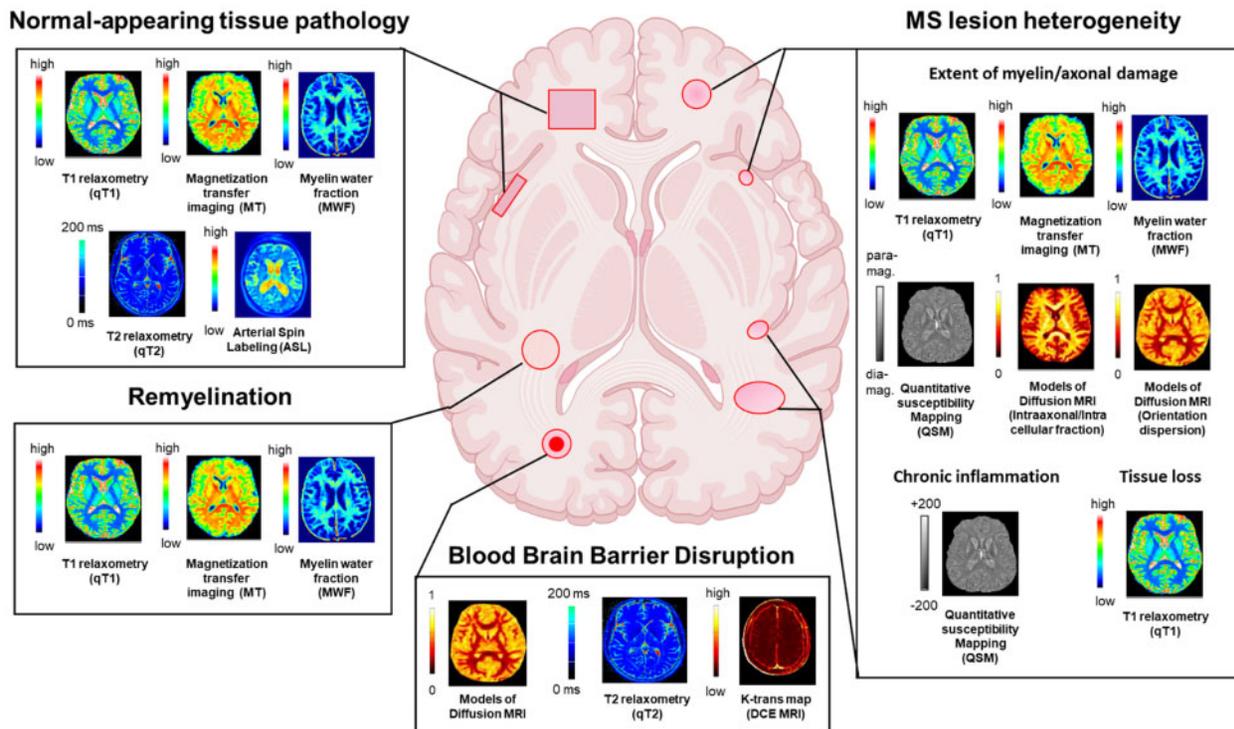


Figure 1 Information provided by quantitative MRI about key features of multiple sclerosis pathology for clinical applications in patients with multiple sclerosis. Quantitative MRI provides information about normal-appearing tissue pathology, multiple sclerosis lesion heterogeneity, remyelination, and blood–brain barrier disruption. dia-mag = diamagnetic; para-mag = paramagnetic.

volume of T_1 -hypointense lesions (also called black holes); and (iv) global/regional volume of tissues (a measure of atrophy). However, T_2 -weighted lesions are not pathologically specific, because they may represent active inflammation (e.g. oedema) as well as demyelination with or without axonal loss. T_1 -hypointense lesions may also result from variable damage to different CNS tissue components (myelin/axon/cells), which cannot be distinguished. Therefore, T_2 -weighted and T_1 -hypointense lesions provide only basic information relating to the histopathological heterogeneity of multiple sclerosis lesions^{5,6} and the different stages of lesion development and repair (e.g. remyelination) that may occur over time. Additionally, brain atrophy reflects only the late-stage results of degenerative phenomena and does not describe the normal-appearing tissue pathology preceding tissue volume loss. In fact, conventional MRI is mostly insensitive to the early and subtle axonal pathology,⁷ alterations in myelin morphology (e.g. myelin blisters⁸) and early-stage dendrite/synapse changes such as those occurring in the hippocampus.⁹

Quantitative MRI techniques

T_1 relaxometry

T_1 relaxometry measures the recovery of longitudinal magnetization of excited spins in a tissue by providing T_1 relaxation time (T_1 -RT) values, which are related to the integrity of micro- and macro-structural components of a tissue¹⁰ (Table 1).

Pathological evidence

Several studies have explored the sensitivity and specificity of T_1 -RT for detecting multiple sclerosis pathology (Table 1). The three major determinants of T_1 changes in the CNS of patients with multiple sclerosis are myelin, iron, and water. While it is possible to model their effect on T_1 -RT,¹¹ it is challenging to disentangle the relative

contributions of myelin, axons, and free water (e.g. oedema in active lesions) to T_1 -RT.¹² Indeed, T_1 -RT highly correlates with both myelin ($r = -0.78$, $P < 0.001$) and axon content ($r = -0.62$, $P < 0.001$) within the normal-appearing white matter and within white matter lesions in the CNS.¹³ Moreover, demyelination, axon loss, and iron loss may all lead to prolonged relaxation times.^{13,14} Interestingly, lesions with longer T_1 -RT are more destructive due to a combination of axonal loss and accumulation of extracellular water.¹⁵ On the other hand, shortening or moderate prolongation of T_1 -RT over time may suggest reparative phenomena such as remyelination and gliosis.¹⁶

Assessment of multiple sclerosis impact and prognostic value

T_1 -RT mapping displays high accuracy for discriminating focal lesions in both white and cortical grey matter in patients with multiple sclerosis.^{17,18} Ultra-high-field (7 T), T_1 -RT mapping can identify cortical focal pathology in cerebral hemispheres¹⁹ and cerebellum²⁰ of patients with multiple sclerosis. In addition, whole-brain T_1 -RT maps at 3 T provide a personalized approach with which to assess the heterogeneity of damage in focal lesions and the extent of diffuse pathology by quantifying the changes in T_1 -RT compared to the normal distribution of T_1 -RT in healthy subjects.²¹ T_1 -RT mapping studies have been used to generate whole-brain assessments of multiple sclerosis disease impact and progression. These studies found that global T_1 -RT increases with disease progression, predominantly in later disease stages and also correlates with brain atrophy.^{22,23} The volume of lesions with very long T_1 -RT (black holes) better correlates with composite clinical functional scores compared with total lesion volume,²⁴ and the decrease over-time in T_1 -RT inside black holes is associated with clinical improvement²⁵ and response to therapy.²⁵ Finally, patterns of T_1 -RT change associated with cognitive impairment can be observed even at early multiple sclerosis stages.²⁶

Table 1 Technical background and pathological specificity and sensitivity of quantitative MRI techniques

Quantitative MRI technique	Contrast mechanism	Measure(s)	Specificity to multiple sclerosis pathology	Sensitivity to multiple sclerosis pathology
qT ₁	Recovery of longitudinal magnetization	T ₁ -RT/R ₁	Low: myelin/axons/cells/macro- and micro molecules/water)	High: (lesions and NAT)
T ₂ relaxometry	Loss of spin Coherence of water pools (myelin layers, intracellular, intra-axonal, extracellular)	T ₂ -RT/R ₂	Low: myelin/axons/cells/water)	High: (lesions and NAT)
MWI	Loss of spin coherence of water molecules trapped in myelin	MWF	High: myelin	High: (lesions and NAT)
MTI	Exchange of magnetization between free protons and macromolecular protons (proteins/lipids)	MTR	Low: myelin/macromolecules (e.g. lipid/protein in biological membranes) extracellular water	High: (lesions and NAT)
DTI	Diffusivity of water proteins (intracellular-extracellular)	MD, RD/AD, FA	Low Highly dependent on tissue structure (e.g. fibre crossing/activated microglia/cells)	High: (lesions and NAT)
Diffusion-based models of microstructure	Modelling of water compartments	Restricted water fraction (CHARMED)	High ODI: neurite dispersion	High: (lesions and NAT, little evidence)
	Modelling of the diffusion magnetic resonance signals	Axon calibre (ACTIVEX) Diffusion Kurtosis ICVF ODI fis	Moderate NDI: myelin and axonal count fis: Neurite and soma	
QSM	Local changes in tissue composition cause frequency shifts (measured by phase images)	Magnetic susceptibility	Low: Influenced by changes in iron/myelin/water content	Moderate: (lesions)
Perfusion MRI				
ASL	Magnetically labelled blood is used as endogenous tracer	CBV CBF	Moderate: linked to mitochondrial energetic failure;	Moderate: (NAT)
DSC	Susceptibility effect of the paramagnetic contrast agent leads to signal loss in T ₂ /T ₂ *-weighted images	MTT K _{trans} V _e V _p	linked to elevated levels of endothelin-1	
DCE	Wash-in, plateau, wash-out of contrast enhancement			

The evaluation of 'specificity' and 'sensitivity' of quantitative MRI measures has been made along two criteria: (i) the strength of correlation between quantitative MRI measures with a given neuropathological feature (specificity); and (ii) the number of neuropathological features measured with quantitative MRI (sensitivity). Based on those criteria, an expert consensus was reached a consensus among the participants of the MAGNIMS workshop (Basel, December 2019) on 'Quantitative MRI towards clinical application in MS'. ASL = arterial spin labelling; CBF = cerebral blood flow; CBV = cerebral blood volume; DCE = dynamic contrast-enhanced; DSC = dynamic susceptibility contrast; fis = soma signal fraction; GM = grey matter; GRASE = gradient and spin echo; ICVF = intracellular volume fraction; K_{trans} = transfer constant; MTI = magnetization transfer imaging; MTT = mean transit time; MWI = myelin water imaging; NDI = neurite density index; ODI = orientation dispersion index; V_e = fractional volume of the extracellular space; V_p = fractional volume of the plasma space.

Technology availability in the clinic

To date, T₁ relaxometry has not been included in clinical multiple sclerosis protocols. There are several reasons for this: (i) numerous approaches have been proposed, exhibiting variable sensitivity to spurious effects such as magnetization transfer (MT), T₂ relaxation, diffusion, and B1 variation; (ii) there is no consensus for the best T₁ mapping sequence^{27,28}; (iii) obtaining high accuracy for T₁ mapping *in vivo* is still challenging²⁹ (e.g. T₁ relaxation in white matter is double-exponential because of magnetization exchange with myelin-bound protons, but most available methods assume single-exponential relaxation); and (iv) the complexity of the T₁ mapping techniques often results in lack of reproducibility³⁰ (Table 2).

Despite challenges with reproducibility of T₁ relaxometry, there are some promising T₁ mapping approaches combining 'clinically

compatible' scan times with high intra- and inter-scanner reproducibility. One of these is the magnetization prepared 2 rapid acquisition gradient echoes (MP2RAGE) sequence,³¹ which has been shown to provide highly reproducible T₁-RT maps (3% coefficient of variability, CV) in a single-vendor, multicentric study³² and in a T₁ phantom study (NIST, National Institute of Standards and Technology) across different 3 T scanners.³³ MP2RAGE T₁ maps provide a promising 'all-in-one' candidate for clinical practice, but achieving this will require manufacturers to collaborate to provide similar acquisitions across different scanners. In addition, clinical cut-offs of pathological changes in MP2RAGE maps still need to be defined. Another interesting approach is synthetic MRI (SyMRI[®]), which simultaneously analyses T₁/T₂ relaxometry and proton density.³⁴ The SyMRI sequence and the software to reconstruct SyMRI maps can be implemented across scanners from all major vendors. Further,

Table 2 Current state of reproducibility and availability for use in humans

Quantitative MRI technique	Inter-scanner reproducibility	Hardware/software availability for clinical use
qT ₁	Moderate High (MP2RAGE)	Limited
T ₂ relaxometry	Moderate	Limited
MWI	High (little evidence)	Limited
MTI	Low/moderate	Limited
DTI	Moderate	Broad
Models of diffusion-based microstructure	Moderate (little evidence)	Limited
QSM	High (little evidence)	Limited
Perfusion MRI (ASL, DSC, DCE)	High	Broad

Reproducibility (inter-scanner and same field strength): high = <5% coefficient of variation (CV); moderate = 5–15% CV; low = > 15% CV in reported studies. ASL = arterial spin labelling; DCE = dynamic contrast-enhanced; DSCE = dynamic susceptibility contrast; MTI = magnetization transfer imaging; qT₁ = quantitative T₁.

quantitative MRI maps can either be generated within a minute of the acquisition or can be installed as a call button on picture archiving and communication systems.³⁵ Moreover, SyMRI showed less than 6% variability in T₁-RT across scanners from different vendors.³⁶

Interestingly, MP2RAGE T₁ mapping has been recently reported in the spinal cord in < 10 min, with a maximum CV of 5%.³⁷

T₂ relaxometry, myelin water fraction, and magnetization transfer imaging

Myelin-sensitive metrics are essential to investigate multiple sclerosis pathophysiology^{38,39} (Fig. 1), perform outcome predictions,^{40,41} and assess therapeutic effects.^{42,43}

Single-component T₂ relaxometry (qT₂) is obtained by fitting a single exponential and provides measures of T₂-RT that are sensitive to global water content in the CNS (intra/extracellular water and myelin water). Nevertheless, since T₂ decay in the CNS tissue is largely multi-exponential, single-component qT₂ is highly dependent on sequence parameters and noise^{44,45} (Table 1).

The distinction of different water pools, including the myelin water pool [e.g. myelin water fraction (MWF)], in the CNS may be achieved by using multi-component T₂ relaxometry⁴⁶ (Table 1). MT imaging exploits the selective saturation of protons bound to macromolecules, including myelin, and reduces their longitudinal magnetization. This renders it possible to create MT saturation images (MTsat) and magnetization transfer ratio (MTR) images providing information about this pool of molecules¹³ (Table 1).

Single-component T₂-RT, MWF, and MTR are also sensitive to the relative presence of extracellular water. For example, their values may be influenced by the presence of oedema^{44,47} (Table 1).

Other techniques, such as rapid estimation of myelin for diagnostic imaging (REMyDI), which is derived from SynMRI, can also be used to assess myelin integrity. REMyDI quantifies myelin by estimating its fast relaxation rate through magnetization exchange and effects on the observable proton pool (i.e. cellular water, free water, and excess parenchymal water partial volumes).^{34,48}

Pathological evidence

Myelin-related measures exhibit different specificity in regard to myelin content and myelin-related pathology in multiple sclerosis.

Post-mortem single-component T₂ relaxometry in the normal-appearing tissue of the cervical spinal cord of patients with multiple sclerosis is highly influenced by demyelination ($r = 0.77$,

$P < 0.001$) but does not seem to be strongly related to axonal damage ($r = -0.44$, $P < 0.001$).⁴⁹ Additionally, qT₂ shows a strong linear correlation with iron concentration in healthy brains ($r^2 = 0.67$, $P < 0.001$).⁵⁰

MWF shows strong correlations with myelin staining in both lesions and normal-appearing tissue in histological specimens of human brain (on average in lesions and normal-appearing tissue: $r^2 = 0.67$, $P < 0.0001$).^{46,51} Whether MWF is also sensitive to accumulation of extracellular iron remains to be demonstrated. A recent post-mortem study attempted to answer this question by imaging brain specimens with two different techniques measuring MWF [Carr Purcell Meiboom Gill (CPMG), and gradient and spin echo (GRASE)], before and after a de-ironing procedure.⁵² This work concluded that both were sensitive to brain iron content; however, this conclusion should be taken with caution, since the applied de-ironing procedure may well have affected the iron content within myelin and, as a consequence, altered myelin properties. Therefore, more studies are warranted to better understand the effect of extracellular iron on quantitative MRI techniques measuring MWF.

MTR, but not MTsat, has been validated by post-mortem studies and shows high correlations with myelin ($r = 0.84$, $P < 0.001$)⁵³ and axon density ($r = 0.83$, $P < 0.0001$)⁵⁴ when lesions and normal-appearing tissue are considered together. In addition, very recent MRI-pathology studies demonstrate that MTR sampled in lesion and non-lesion tissue in multiple sclerosis brains is weakly associated not only with myelin density [coefficient (95% confidence interval, CI): 0.31 (0.07 to 0.55), $P = 0.01$], but also with greater numbers of astrocytes [coefficient (95% CI): 0.51 (0.02 to 1), $P = 0.04$] and damaged mitochondria [coefficient (95% CI): 0.53 (-0.95 to -0.12), $P = 0.01$].⁵⁵

Amongst the most recent myelin-sensitive approaches, REMyDI myelin quantification was shown to weakly correlate with both proteolipid protein (PLP) immunostaining and Luxol fast blue staining in multiple sclerosis lesions but not with the same staining in normal-appearing white matter (multiple sclerosis lesions: $0.02 < r < 0.48$, $P < 0.001$).⁵⁶

Assessment of multiple sclerosis impact and prognostic value

In patients with short disease duration (i.e. <6 years), T₂-RT values were increased in normal-appearing white matter^{57–59} and normal-appearing grey matter⁶⁰ as compared to healthy controls. T₂-RT in combination with diffusion MRI appeared to be sensitive also to extracellular water accumulation due to blood-brain barrier disruption in gadolinium-positive lesions, which could be identified with 85% accuracy using these two measures.⁶¹ Similarly, MWF

increased over 6 months following the appearance of enhancing lesions, as expected in repairing tissue.⁶² Additionally, MWF was reported to be lower in lesions with paramagnetic rims compared with rim-negative lesions,⁶³ showing the more pronounced demyelination in the former lesion type. Also, MWF moderately decreased over time (–8%) in the normal-appearing white matter of a small group of patients followed over an average period of 5 years. Global MWF was likewise abnormal in the brain and cervical spinal cord of patients with primary progressive multiple sclerosis compared to controls,⁶⁴ and, when followed for 2 years, decreased at the C2–C3 spinal cord level⁶⁵; moreover, cervical spinal cord MWF showed an association with disability.⁶⁵

MTR changes in normal-appearing white matter preceded the appearance of gadolinium-enhancing lesions in patients with multiple sclerosis⁶⁶ and recovered following the acute phase,⁶⁷ especially in treated patients.⁶⁸ MTR was significantly lower in hypointense lesions as compared with isointense lesions on T₁-weighted images at the time of initial enhancement.⁶⁹ For lesions that changed from hypointense to isointense, MTR increased significantly during 6 months of follow-up.⁶⁹ Intralesional MTR showed longitudinal changes consistent with demyelination and remyelination in different regions of active lesions in the 3 years following treatment.⁷⁰ MTR also appeared to be lower in outer compared to inner cortical layers in the brain, and in the subpial region compared to the central region in the spinal cord⁷¹; this may be consistent with differences in myelin content, but may also, at least in part, be due to partial volume effects. The lowest outer cortical MTR was seen in secondary progressive multiple sclerosis and is consistent with more extensive outer cortical (including subpial) pathology.⁷² MTR abnormalities in the subpial region, in both brain and spinal cord, occurred early in the course of multiple sclerosis and were more marked in patients with a progressive disease course.⁷³ As for the spinal cord, lower MTR values were found in the cervical cord of patients with RRMS^{71,74} and primary progressive multiple sclerosis⁷⁴ compared to controls, which further decreased over 5 years follow-up.⁷⁴

In contrast to MTR, MTsat has been evaluated in patients with multiple sclerosis in a few studies, revealing its sensitivity to normal-appearing white/grey matter abnormalities^{75,76} and multiple sclerosis lesions.^{76,77} A proper comparison between the sensitivity of MTsat and MTR to multiple sclerosis pathology has still to be performed, but there is preliminary evidence that MTsat in the cervical spinal cord better correlates with disability than MTR.⁷⁸

Myelin maps provided by REMyDI showed increased myelin loss in normal-appearing white matter of patients with multiple sclerosis compared to controls, which correlated with baseline cognitive and physical disability.⁵⁶ Longitudinally, MWF correlated with follow-up physical disability, even after adjusting for baseline disability.⁵⁶

Technology availability in the clinic

Up to now, MTR/MT sat, qT2, and MWF have not been available for clinical use (Table 2). However, there are now some sequences that provide qT2 and MWF maps in 3–6 min, a time that may be compatible with clinical protocols.^{79–82} Similarly, for the cervical spinal cord, fast acquisitions for MWF begin to be available.⁸³

There are also several sequences available to perform MT imaging, but none can provide the reconstruction of MTR maps in a clinical setting. Interestingly, the comparison of different myelin-sensitive methods (GRASE- and mcDESPOt-MWF, qT1, and MTR) indicates that the type of sequence needs to be chosen according to the purpose of its application. For example, the GRASE sequence should be used when the greatest confidence is required for assessing changes specific to myelin.³⁹ If sensitivity to lesion and

normal-appearing white matter pathology is the priority, then mcDESPOt-MWF and qT1 are the most sensitive approaches.³⁹

Although achieving reproducible MTR measurement has traditionally been challenging, by carefully controlling the sources of technical variability (protocols, type of coil for MT saturation, and B1 inhomogeneities), it is feasible to obtain inter-scanner MTR in healthy controls, the variability of which lies within the mean inter-subject variability.^{84,85} Also, MTsat exhibits moderate variations [CV intra-scanner 7–12%; CV inter-scanner: 15.7% (<5% if MT harmonization is performed)⁸⁶].

Intra-scanner reproducibility of MWF in white matter is quite high ($r = 0.95–0.99$),^{81,82,87} and the same holds true in small areas simulating multiple sclerosis lesions ($r = 0.89$ ⁶²). Slightly lower, but still very good, is the inter-scanner and inter-vendor reproducibility for some myelin water imaging sequences in white matter ($r = 0.91$, CV < 3%⁸⁸) although more studies are required to elucidate this aspect also for other ‘clinically compatible’ MWI sequences.

An initial assessment of reproducibility of myelin maps as provided by SyMRI, including REMyDI, showed moderate reproducibility across vendors ($\rho = 0.89$).⁷⁶

In the spinal cord, measurement errors for MTR and MTsat remain very large,⁸⁹ and more data are required to understand the intra- and inter-scanner reproducibility of fast spinal MWF acquisitions.

Diffusion microstructure

Diffusion MRI probes CNS tissue integrity using metrics derived from modelling signal changes associated with the diffusion of water molecules in tissue, which can characterize cellular compartments of brain tissue within multiple sclerosis lesions, normal-appearing white matter, and normal-appearing grey matter (Table 1 and Fig. 1).

Diffusion tensor imaging (DTI) is a technique widely available in clinical research and clinical practice. DTI-derived metrics [fractional anisotropy (FA), radial/axial diffusivity (RD/AD), and mean diffusivity (MD)/apparent diffusion coefficient (ADC)] have been used for many years to assess the CNS tissue integrity in both regions of interest and along specific white matter tracts.^{90–93} Beyond DTI, several mathematical models and computational approaches have attempted to decode the information contained in diffusion-weighted signals to retrieve specific features of tissue microstructure by: (i) modelling the tissue (e.g. tissue geometry and water dispersion) and associated signals; or (ii) computationally exploring the magnetic resonance signal (e.g. assessing signal behaviour with minimal or no underlying geometrical assumptions). Some models have attempted to separate different water compartments (extracellular, intracellular, and other) within CNS tissue.⁹⁴ These approaches normally require diffusion acquisitions that are more complex than the ones clinically used for DTI, encompassing multiple b-values and sampling the signal in numerous directions. Extensive work comparing different diffusion-weighted imaging models and their ability to explain acquired data demonstrated that on average, and with the methods tested, tissue models tend to explain diffusion-weighted signal behaviour better than do signal models.⁹⁵

The composite hindered and restricted model of diffusion (CHARMED) separates the intra- and extracellular water compartments and generates maps of the restricted water fraction (FR), a proxy for axon density.⁹⁶ The CHARMED framework has been extended to account for different axonal diameters, providing the opportunity to map the distribution of axon diameters within the brain using AxCaliber^{97,98} or ActiveAx⁹⁹ frameworks. Another method is diffusion kurtosis imaging (DKI), which aims to provide a more accurate model of diffusion-weighted signal changes capable of capturing non-Gaussian diffusion behaviour as a reflection

of tissue heterogeneity.¹⁰⁰ Diffusion-based spectrum imaging (DBSI) models the diffusion signal as a linear combination of anisotropic diffusion tensors reflecting fibres, which are predominantly axon fibres in white matter, and a spectrum of isotropic diffusion tensors that encompass cells, oedema, and CSF.^{101,102}

Neurite orientation dispersion and density imaging (NODDI) is a three-compartment tissue model providing metrics to measure the intracellular volume fraction (ICVF) or neurite density index (NDI) and the orientation dispersion index (ODI), which describe intracellular diffusion in terms of neurite density and the degree of fibre dispersion of neuritis, respectively.¹⁰³ Soma and neurite density imaging (SANDI) is another tissue-model aimed at further distinguishing the intracellular space by separately modelling the intra-neurite and intra-soma spaces.¹⁰⁴

Q-space imaging may be applied to study the microstructural changes in white matter by estimating the water diffusion function, the probability density function (PDF), also called mean apparent propagator (MAP)¹⁰⁵ or ensemble average propagator (EAP).^{106,107}

Pathological evidence

Post-mortem studies showed that DTI-derived FA and MD in normal-appearing white matter correlate to myelin content ($r = -0.79$ and $r = 0.68$, $P < 0.001$ for both) and to a lesser degree axon count ($r = -0.7$ and $r = 0.66$, $P < 0.001$ for both).¹⁰⁸ Also, in the cortex of non-neurological subjects and patients with multiple sclerosis, FA values strongly relate to axon density [β (95% CI) = 1.56 (0.69 to 2.44) and β (95% CI) = 0.93 (0.45 to 1.42), $P < 0.05$ for both] but not to myelin, glia, and total cell density.¹⁰⁹ However, these results should be taken with caution as the relationship between DTI parameters and myelin/axon content decreases in lesions and varies in regions of complex microstructure (e.g. crossing fibres^{110–113}).

Post-mortem validation of microstructural features derived from biophysical diffusion models has been performed for some models. However, very few of these models have been evaluated in multiple sclerosis tissue specimens. AxCaliber showed a very high agreement between the estimated axon diameter distribution in various nerve samples and axon diameter histograms on histology images ($r = 0.98$ for optic nerve and $r = 0.86$ for sciatic nerve⁹⁷). ActiveAx maps of axon diameter and density indices exhibited a similar distribution pattern to those observed histopathologically in the corpus callosum and corticospinal tract.⁹⁹ NODDI ODI has been reported to correspond well with histological measures of neurite orientation dispersion in brain and spinal cord (controls: $r = 0.84$; $P < 0.001$; multiple sclerosis: $r = 0.60$; $P = 0.001$), whereas NODDI NDI showed good correlation with myelin ($r = 0.74$; $P < 0.001$) and moderate correlation with histology-derived neurofilament density measures ($r = 0.56$; $P = 0.002$).^{114,115} In post-mortem specimens of multiple sclerosis lesions in the spinal cord, lower NDI and increased ODI were observed compared to non-lesion tissue.¹¹⁴ Also, in the same specimens, NDI was reported to be sensitive to myelin and axon count.¹¹⁴ As to DBSI, the study of one biopsied tumefactive multiple sclerosis lesion showed that DBSI-derived AD better detected axonal loss than DTI¹¹⁶ but no correlation between DBSI parameters and axonal density was reported.

Other emerging microstructural diffusion models still require histopathological validation in healthy controls and multiple sclerosis brain specimens.

Assessment of multiple sclerosis impact and prognostic value

Even though DTI measures provide only a coarse approximation of CNS tissue properties, they have been extensively used in multiple

sclerosis research studies. Increases in MD/ADC have been reported up to 6 weeks before contrast enhancement,^{117,118} and MD in enhancing lesions has been shown to be much lower than in non-enhancing lesions.¹¹⁹ Increased MD in acute multiple sclerosis lesions also appeared to predict risk of developing persistent black holes.¹²⁰ Furthermore, DTI measures along white matter tracts showed a progressive increase in MD in patients with no evidence of clinical or radiological disease-activity,¹²¹ and a decrease in RD as well as an increase in AD in progressive patients with multiple sclerosis.⁹¹ In addition, the peak width of skeletonized MD appeared to be higher in RRpatients with multiple sclerosis compared to controls.¹²² Recent DTI studies in the cervical spinal cord have reported increased RD and reduced FA in RRMS with acute spinal cord involvement, when compared with healthy controls, and in SPMS, when compared with clinically stable RRMS.^{123,124}

DTI metrics in the brain can also predict disability progression¹²⁵ and cognition,¹²⁶ especially in combination with clinical variables.¹²⁷ Likewise, RD in the optic nerve is inversely related to visual acuity in patients with multiple sclerosis¹²⁸ and has been shown—together with FA—to correlate with clinical disability in patients with spinal cord lesions.^{124,129} Furthermore, baseline RD in the cervical spinal cord over a 6-month period during an acute cord relapse correlates with recovery at 6 months.¹³⁰ Also, FA and MD change over time in patients with multiple sclerosis, but those measure do not seem to relate to changes in disability.¹³¹

DKI measures (e.g. mean kurtosis) are affected in patients with multiple sclerosis compared to controls¹³² and are related to patient's disability.¹³³ DBSI-derived measures in white matter lesions and in the corpus callosum distinguished clinical multiple sclerosis subtypes with moderate accuracy¹³⁴ and also different types of multiple sclerosis lesions.¹³⁵

The CHARMED-derived restricted water fraction is decreased in early multiple sclerosis, both in lesions and in normal-appearing white matter,¹³⁶ and decreases over time in lesions and normal-appearing white matter.¹³⁷ NODDI shows a lower neurite density (NDI) together with a higher neurite orientation dispersion (ODI) in normal-appearing white matter and in multiple sclerosis lesions compared with healthy white matter, both in the brain^{137–139} and in the cervical spinal cord.^{114,140,141} NODDI abnormalities are more pronounced in patients with SPMS than RRMS.¹³⁹ Additionally, NODDI measures better correlate with disability and cognitive/motor function in patients with multiple sclerosis than do standard DTI metrics.¹³⁹

Last, q-space imaging (QSI) perpendicular diffusivity is higher and parallel diffusivity lower in the cervical spinal cord of progressive PPMS compared to healthy subjects,¹⁴² and those changes become more evident over time.¹⁴³ Also, an increase in cord QSI indices of perpendicular diffusivity is associated with disability worsening over 3 years in PPMS.¹⁴³

Technology availability in the clinic

Currently, DTI protocols are available for most clinical scanners and could be used in clinical practice, although DTI measures are more often used for comparisons between patient groups in research studies than for management of individual patients. More studies are needed to better understand the clinical validity of DTI-derived metrics for patients with multiple sclerosis.¹¹⁰

Reproducibility of DTI metrics has been assessed in numerous studies, which showed that FA has an intra-scanner coefficient of variation $< 3\%$ and MD 0–7%, whereas the inter-scanner coefficient of variability for both FA and MD is reported

to be $\leq 5\%$.^{144–150} Nevertheless, further studies should assess DTI reproducibility in multiple sclerosis cohorts.

Microstructural models applied to multi-shell diffusion data are far from being ready for clinical adoption, and only few reproducibility studies have been performed. Among those, some works reported an inter-vendor reproducibility of NODDI ranging from 2.3% to 14% and an intra-scanner reproducibility $\leq 4\%$.^{151,152} More works are required to understand the potential clinical role of microstructural metrics and their reproducibility across scanners and vendors.

As for the spinal cord, a recent investigation of the reproducibility of DTI-derived measures at C1–C6 between different sites has shown the feasibility of multicentre spinal cord DTI, when sequence parameters are homogenized across sites and vendors.¹⁵³

Quantitative susceptibility mapping

Quantitative susceptibility mapping (QSM) encompasses imaging methods by which the absolute concentrations of iron, calcium, myelin, and other substances may be measured in tissues based on their changes in local magnetic susceptibility (Table 1). In particular, the magnetic susceptibility is calculated from local frequency shifts in the MRI signal of a gradient echo sequence (as obtained from the phase images) through deconvolution with a dipole kernel. Several methods have been proposed to solve this ill-posed inverse problem.¹⁵⁴ QSM maps can quantify paramagnetic trace elements, such as iron in ferritin, deoxygenated-heme in the blood, and diamagnetic calcium. In addition, myelin¹⁵⁵ and the microstructural anisotropy of white matter^{156–159} can also induce local shifts of the magnetic susceptibility because of the diamagnetism of proteins and lipids.¹⁶⁰ QSM also provides an improved contrast-to-noise ratio for certain tissues and structures compared with T_2^* -weighted magnitude images. However, because of phase filtering, QSM does not provide absolute susceptibility values, and, therefore, QSM is computed in relationship to a reference region.¹⁶¹ Moreover, the magnetic susceptibility in white matter is a tensor (i.e. it depends on fibre orientation with respect to the main magnetic field, B_0), which can make the interpretation of susceptibility changes in white matter challenging.¹⁶²

Pathological evidence

A post-mortem study before and after brain fixation at 7 T showed that QSM is positively related to ferritin iron content ($r = 0.76$) and negatively related to myelin content ($r = -0.35$), which indicates a paramagnetic effect of iron and a diamagnetic effect of myelin on tissue magnetic susceptibility.¹⁶³ In multiple sclerosis brain samples, QSM identifies iron accumulation in microglia and macrophages surrounding chronic active and smoldering lesions^{164,165} as well as active myelin digestion during lesion formation.¹⁶⁶

Assessment of multiple sclerosis impact and prognostic value

QSM reveals magnetic properties sensitive to iron and myelin, and thus can capture specific characteristics of multiple sclerosis lesions (Fig. 1). Longitudinal QSM measurements in patients with multiple sclerosis have shown an initial large rise in magnetic susceptibility occurring within weeks in active lesions, and a subsequent increase that occurs for months.¹⁶⁷ The former has been attributed to myelin digestion and the latter to the removal of the myelin debris within macrophages and the release of iron.¹⁶⁷ In addition, lesions with higher susceptibility at the border and larger volume maintained a high magnetic susceptibility value for a number of years,¹⁶⁷ a finding confirmed by susceptibility-weighted imaging (SWI) at ultrahigh field.¹⁶⁴ These lesions are particularly interesting as they are thought

to contain smouldering inflammation and are associated with rapid clinical progression.¹⁶⁸ Magnetic susceptibility values in the basal ganglia were higher in patients with clinically isolated syndrome and multiple sclerosis than in control subjects.¹⁵⁴ Furthermore, this increased iron deposition in the basal ganglia, measured by QSM at 7 T, correlated with cognitive measures of inhibitory control in patients with multiple sclerosis.¹⁶⁹

Magnetic susceptibility values from QSM maps showed promise in detecting enhancing lesions without the use of gadolinium.¹⁶⁷ Finally, QSM is also sensitive to the oxygenation state of blood, thereby allowing calculation of the oxygen extraction fraction. Thus, patients with multiple sclerosis were found to exhibit lower oxygen extraction fraction than controls, which is possibly related to mitochondrial dysfunction.¹⁷⁰

Technology availability in the clinic

It is relatively easy to collect data that may be used to reconstruct QSM maps on clinical MRI scanners. In fact, many clinical protocols are already applying 2D or 3D GRE (gradient echo) sequences to obtain T_2^* -weighted or SWI, and these protocols may also be used for QSM if phase images are available. Currently, the main hurdle for the broad translation of QSM into clinics is that MRI vendors have yet to implement the necessary algorithms on their commercial scanners. In addition, offline reconstruction of QSM maps is laborious and not easy to implement in routine clinical practice. Also, there is currently no consensus about which algorithm is best for QSM reconstruction. Most current QSM approaches suffer from over-smoothing and loss of conspicuity of fine features, as the methods are primarily optimized to minimize error metrics, not improve image quality.¹⁷¹

Brain QSM measurements performed by using the same algorithm in different magnetic resonance scanners exhibit: (i) excellent inter- and intra-scanner reproducibility for healthy subjects ($r = 0.99$ and $r = 0.98$, respectively)¹⁷²; (ii) very high intra-scanner reproducibility for patients with multiple sclerosis ($r = 0.97$)¹⁷²; and consistently high intra- and inter-scanner reproducibility in phantoms with different gadolinium concentrations.¹⁷³

To date, QSM has not been developed for spinal cord imaging.

Perfusion imaging

Blood perfusion in the brain can be assessed using a tracer injection (e.g. gadolinium-based contrast agents) during the MRI acquisition of: (i) a T_2^* -weighted dynamic susceptibility contrast (DSC) sequence, which may provide relevant parameters in patients with multiple sclerosis such as: cerebral blood flow, cerebral blood volume, and mean transit time; or (ii) a T_1 -weighted dynamic contrast enhancement (DCE) sequence able to measure the volume transfer constant K_{trans} , which is a measure of permeability between blood plasma and tissue extravascular spaces and of plasma blood flow and capillary surface area. Alternatively, arterial spin labelling (ASL), a technique which does not require intravenous administration of gadolinium-based contrast agents, uses magnetically labelled blood as the intrinsic contrast agent, most commonly measuring cerebral blood flow and volume.

Pathological evidence

Chronic hypoperfusion induces mitochondrial dysfunction leading to energetic failure and oxidative stress, which are increasingly recognized as crucial factors associated with axonal degeneration in multiple sclerosis.¹⁷⁴ Furthermore, generalized microstructural damage in normal-appearing white matter could be associated with elevated levels of endothelin-1, a vasospastic peptide.¹⁷⁵

Assessment of multiple sclerosis impact and prognostic value

Changes in local perfusion are known to precede the initial blood-brain barrier breakdown and T₂-weighted lesion appearance by several weeks.¹⁷⁶ In general, lesions tend to develop preferentially in hypoperfused brain areas,¹⁷⁷ but the relation of hypoperfusion to T₂-weighted lesion load is controversial.^{178–180} Generalized hypoperfusion in normal-appearing white matter is correlated with microstructural damage in the brain parenchyma.¹⁷⁴ Brain perfusion is generally reduced in chronic disease phases,^{181,182} is correlated with diffuse axon degeneration,¹⁷⁴ and precedes atrophy development.¹⁸³ Gliosis also induces less metabolic demand and results in decreased perfusion.¹⁷⁶ Reductions in cerebral blood volume and cerebral blood flow in multiple sclerosis are associated with worsening of physical disability¹⁸⁴ and have been widely reported to correlate with disability and composite functional scores.^{181,182,185–187} Correlations with the mean transit time are still controversial, as this parameter is not consistently altered in multiple sclerosis as it is in other conditions such as stroke.^{185–187} A multitude of studies have consistently described the correlation between cognitive decline and reduced perfusion parameters,^{180,188–191} as well as with fatigue in multiple sclerosis.^{188,190,192} Hypercapnic perfusion experiments showed impaired dilatory capacity of cerebral arterioles in multiple sclerosis in response to vasomotor stimulations.¹⁹³

Technology availability in the clinic

In general, perfusion data in multiple sclerosis may be hampered by sensitivity to artefacts, dependency on haematocrit, and the lack of absolute quantification, which may render difficult the interpretation and comparison of data acquired at different time points or across scanners.^{194–196} Standardization of protocols and analyses is currently under development.

Towards clinical application and clinical decision support with quantitative MRI

In theory, quantitative MRI techniques that measure accurately and reproducibly a biologically specific signal correlating with, or

predictive of, clinical outcomes, are ideal candidates for clinical applications. In practice, such techniques do not exist yet (Tables 1 and 2). Among the currently available quantitative MRI approaches, those achieving high accuracy (which usually comes at the cost of reproducibility) are extremely appealing for research investigating the underlying tissue changes; on the other end, those providing high reproducibility, acceptable accuracy, and good correlation with clinical measures, despite not comprehensively explaining clinical function or disability, may be useful for the management of people with multiple sclerosis or in assessing therapies for patients with multiple sclerosis.

Currently, quantitative MRI techniques lack some of the development steps necessary to achieve clinical maturity (Fig. 2). These include clinical availability of acquisition methods and tools to reconstruct parametric maps; methods to extract quantitative information from parametric maps; and normative values and pathological cut-offs (Fig. 2). Additionally, the clinical value of quantitative maps needs to be compared with existing diagnostic and prognostic criteria in the clinical setting. In this context, qT1 (e.g. MP2RAGE and SyMRI) and, to some extent, myelin imaging (e.g. SyMRI) appear to be the most technically advanced and ready for use in studies aimed at providing methods to extract quantitative information, either through brain and spinal cord atlases^{197–199} or comparison of single subjects to large cohorts of healthy cohorts²¹ (Fig. 2). Although brain myelin imaging may be ready for clinical adoption, we still lack software solutions that can provide clinicians with valuable information related to the state of damage or repair of the underlying tissue. QSM and diffusion-based methods providing microstructural information warrant further technological development, and their reproducibility must be assessed in a multicentre setting. Finally, while perfusion MRI may be considered to assess blood-brain barrier permeability and for monitoring disease progression, more studies are needed to provide evidence of its clinical value in multiple sclerosis.

Another important consideration is whether specific quantitative MRI methods are better suited for the characterization of specific multiple sclerosis disease subtypes, assessment of disease progression, and evaluation of therapy response. The data presented in this review suggest that, currently, T₁ relaxometry and QSM may be most suitable for multiple sclerosis stratification by

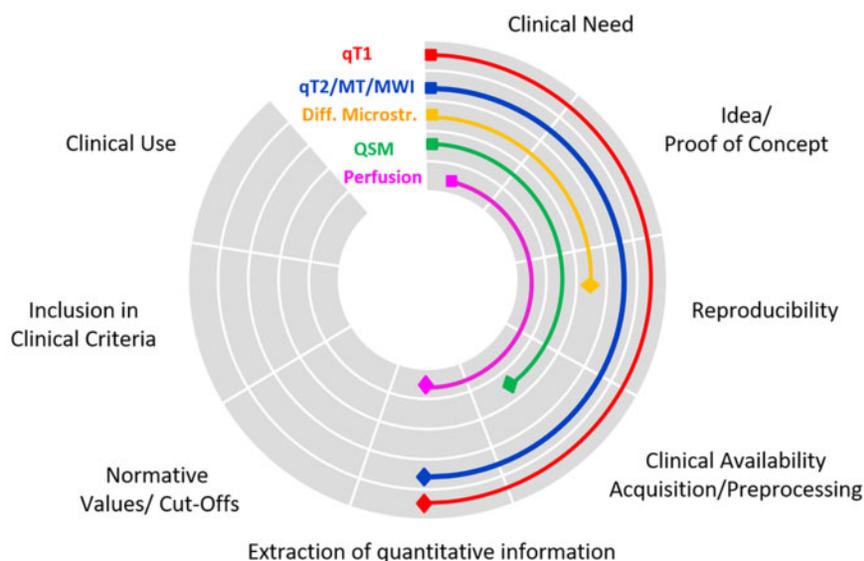


Figure 2 Clinical maturity of the main quantitative MRI approaches. Schematic representation of quantitative MRI current development stages towards clinical use.

contributing to the identification of lesions associated with more extensive tissue damage and to the differentiation of acute versus chronic inflammatory lesions. Myelin-sensitive quantitative MRI techniques [MTR/MTsat, myelin water imaging (MWI), and T_2 relaxometry] and diffusion-microstructure MRI measurements may be most appropriate for assessing clinical progression through the characterization of normal-appearing tissue abnormalities, and may also be used to evaluate therapy effects on specific CNS tissue components (e.g. myelin and axons). As quantitative MRI methods become better standardized, further studies will be required to define their role in the management of patients with multiple sclerosis.

Besides cerebral imaging, quantitative MRI (e.g. DTI, MTR/MTsat, and MWI) also holds promise for imaging of the spinal cord, but both additional software (for localization, gating, and motion compensation) and hardware development (e.g. multi-channel phased-array coils) are required to pave the path towards application of spinal quantitative MRI for multiple sclerosis management.

In summary, quantitative MRI has the potential to provide information that can improve patient stratification, assessment of therapy response, and evaluation of subclinical disease progression. Whether these techniques should be embedded in clinical routines or selected for targeted implementations and studies within the clinical arena is still to be determined. Future work should be targeted at improving quantitative MRI clinical maturity through multicentre collaborations.

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Appendix 1

Authors are members of the MAGNIMS network (Magnetic Resonance Imaging in multiple sclerosis; <https://www.magnims.com>).

eu/), which is a group of European clinicians and scientists with an interest in undertaking collaborative studies using MRI methods in multiple sclerosis, independent of any other organization and is run by a steering committee whose members are: F. Barkhof (Amsterdam), N. de Stefano (Siena), J. Sastre-Garriga (Barcelona), O. Ciccarelli (London), C. Enzinger (Graz, Co-Chair), M. Filippi (Milan), Claudio Gasperini (Rome), L. Kappos (Basel), J. Palace (Oxford), H. Vrenken (Amsterdam), À. Rovira (Barcelona), M.A. Rocca (Milan, Co-Chair), and T. Yousry (London).

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